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INVESTIGATION OF SINGLE-SCAN TV RADIOGRAPHY
FOR THE ACQUISITION OF DYNAMIC PHYSIOLOGIC
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DEVELOPMENT AND INVESTIGATION OF SINGLE-SCAN TV RADIOGRAPHY
FOR THE ACQUISITION OF DYNAMIC PHYSIOLOGIC DATA

Semi-annual Report
May 1, 1975 through October 31, 1975

National Aeronautics and Space Administration
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INTRODUCTION

This period has produced a significant breakthrough in our knowledge of the capabilities for imaging and extraction of quantitative data for systems of the type which have been under investigation. These capabilities are sufficiently greater than previously suspected. They have allowed us to produce computerized axial tomograms from fluoroscopic input images stored on video disc which equal or exceed those produced by the EMI scanner, both in differential contrast sensitivity and in resolution. This development should result in a significantly lower-cost device and a dramatic reduction in patient dose.

A new type of light amplifier for large flat screen fluoroscopy is being investigated which will decrease both its size and weight.

The work on organ contouring is being extended to yield volumes. This is a simple extension since the fluoroscopic image contains density (gray scale) information which can be translated as tissue thickness, integrated, thereby yielding accurate volume data in an on-line situation.

Most approaches to image enhancement have been through the use of the digital computer. This is expensive, time consuming, and therefore not applicable to on-line systems. We have developed a number of devices for analog image processing of video signals, operating on-line in real time, and with simple selection mechanisms. The results show that this approach is feasible and produces a great improvement in image quality which should make diagnostic error significantly lower. These are all low cost devices, small and light in weight, thereby making them usable in a space environment, on the Ames centrifuge, and in a typical clinical situation.

RESEARCH SUMMARY

I. Measurement of Organ Volume

All aspects of our data processing utilize the density (as represented by gray scale values) variations from which the fluoroscopic image is composed. In addition these are stored in the computer and can therefore be used for any number of successive operations. An example of this is the extension of our contouring routine¹ to yield volumes of the enclosed area.

1. Newell, J.D., Keller, R.A., and Baily, N.A., Simple method for the generation of organ and vessel contours from roentgenographic or fluoroscopic images. Med. Phys. 2:73-75 (1975).

This procedure essentially consists of subdividing the enclosed area into a matrix sufficiently fine, recording an associated gray level for that particular matrix element, and then translating this level into the appropriate pathlength. Calibration in terms of a known absorber is all that is required. Summation of these individual columns yields the volume directly. In cardiac investigations this will yield absolute volume at any portion of the cardiac cycle desired, stroke volume, ejection fraction, and any other parameter associated with ventricular volume. Details of the calibration (on-line) are now being investigated.

II. Use of a Digitized X-Ray Fluoroscopic System On The NASA Ames Centrifuges.

The flat screen, x-ray fluoroscopic system installed performed extremely well during its first actual trial during an animal run. Both tape and disc recording were accomplished. The results were so good that bi-plane systems will be designed for use on the centrifuge and in the laboratory. Although the silicon intensified vidicon camera gives excellent results, it requires a higher x-ray flux level than is desirable for humans. It also has the typical blooming evident at high density gradient boundaries associated with other tubes such as plumbicons. However, the feasibility of this type of system for centrifuge or space applications has certainly been established.

III. Large Flat Screen Fluoroscopic Systems.

The various applications and performance characteristics of low light level TV, large flat input screen, fluoroscopic systems has been summarized previously.²

2. Baily, N.A., Development and Investigation of Single-Scan TV Radiography For The Acquisition of Dynamic Physiologic Data. Semi-annual Report, Nov. 1, 1974 through April 30, 1975, NASA Res. Grant No. NGR 05-009-257.

A recent development in the technology of light amplifiers, the Mullard channel plate image tube, appears to be a major advance in the state of the art in this field. Incorporation of this tube, as a replacement for the 3-stage electrostatic light amplifier

previously used in the fluoroscopic system will significantly reduce size and weight and also increase its ruggedness. A system using this device is now under construction. It will be evaluated for sensitivity, resolution, and general image quality.

IV. Response of Fluoroscopic-TV Chains to Incident X-Ray Intensity.

It is a common misunderstanding and assumption that the video output signal in these systems is linear with screen input x-ray flux. In all of our work with such systems we have found that the video output is proportional to the product of the linear absorption coefficient and the absorber thickness.³ We have completed a comprehensive study of three types of systems in use either clinically or in physiological research. A paper was presented at the 1975 Annual Meeting of the Association of University Radiologists and a paper accepted for publication in Investigative Radiology. A preprint is included as Appendix A.

3. Baily, N.A., and Crepeau, R.L., The Capability of Fluoroscopic Systems To Determine Differential Roentgen-Ray Absorption. Radiology 115:439-445 (1975).

V. Computerized Reconstruction of Transverse Body Sections From Fluoroscopic Images.

During this period we have presented two papers,^{4,5} and submitted three others for publication,^{6,7,8} each describing in detail the work reported in the last semi-annual report.²

4., 6. Baily, N.A., Acquisition of Quantitative Physiological Data and Computerized Image Reconstruction Using a Single Scan TV System. Int. Conf. on Cardiovascular Imaging and Image Processing, Stanford Univ., 1975. Proc. of the Conf. on Imaging and Image Processing-Theory and Practice, Proc. of the S.P.I.E., 72. (In Press)

5., 7. Kak, A.C., Jakowatz, C.V., Jr., and Baily, N.A., Computerized Tomography Using Video Recorded Fluoroscopic Images. Topical Meeting on Image Processing for 2-D and 3-D Reconstructions from Projections, Stanford Univ., 1975, Proc. of the Topical Meeting on Image Processing for 2-D and 3-D Reconstructions from Projections: Theory and Practice in Medicine and the Physical Sciences. (In Press)

8. Kak, A.C., Jakowatz, C.V., Jr., Baily, N.A., and Keller, R.A., Computerized Tomography Using Video Recorded Fluoroscopic Images. Submitted to I.E.E.E., Trans. on Biomedical Eng.

In view of the excellent results achieved to date a special phantom was constructed and detailed data on resolution and contrast detectability were obtained. A paper included as Appendix B has been submitted to Investigative Radiology.

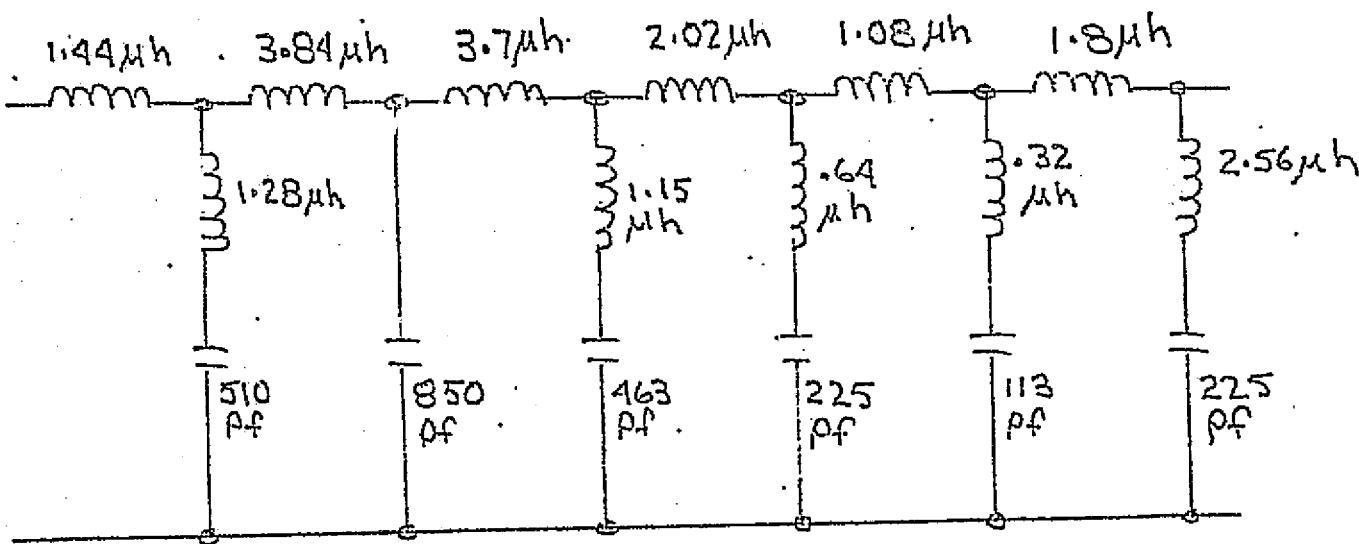


FIG. 1

VI. Enhancement of Fluoroscopic and Roentgenographic Images Using Analog Electronics.

The enhancement of images through the use of digital computers has received wide attention. This method has a number of deficiencies for use in radiology. First, the complexity of the radiologic image is so great that any single computer program tends to enhance certain objects at the expense of others. Control of such programs by the radiologist, cardiologist, etc. is very difficult. Second, processing is expensive. Third, the end result is very much delayed.

A more efficacious route is the use of analog electronic circuitry for processing of the video signal. These can be used directly in fluoroscopy or roentgenograms may be processed from a view-box display. Another advantage is that processors can be used singly or in combination. We have developed and built a number of such devices.

(a) Sharp cut-off, low-pass filter.

One of the major degraders of a video image is electronic noise. In many fluoroscopic and roentgenologic images, frequencies greater than 5 MHz do not contribute substantially to the diagnostic information. However, the noise components having frequencies above this value do degrade the image. In addition all analog processors are more effective in their operation when noise is removed by pre-filtering. As a starting point we therefore have designed and built a sharp cut-off, low-pass filter having a cut-off frequency of 5 MHz. The circuit is shown in Fig. 1. This filter will pass unattenuated all frequencies less than 5 MHz, and remove practically all those above this value. This bandwidth will allow use of a total of 250 line pairs or approximately 20 line pairs/mm at the target. For a typical 9" intensifier this will yield slightly more than 2 line pairs/mm at the input screen. This is slightly greater than the present generation of recording equipment is capable of imaging.

(b) Four-Way Aperture Equalization.

This device takes its name from a similar device used for scanning beam corrections in all modern color television cameras and increases sharpness by reducing optical aperture defects due to the use of a finite number of scan lines, scanning beam size, etc. In our system, an all directional second derivative signal (detail signal) is produced by the special circuitry which produces a similar signal for vertical enhancement in addition to the usual horizontal enhancement.

SYNC STRIPPER

CLAMP PULSE

SYNC OUT

+5V

VIDEO IN

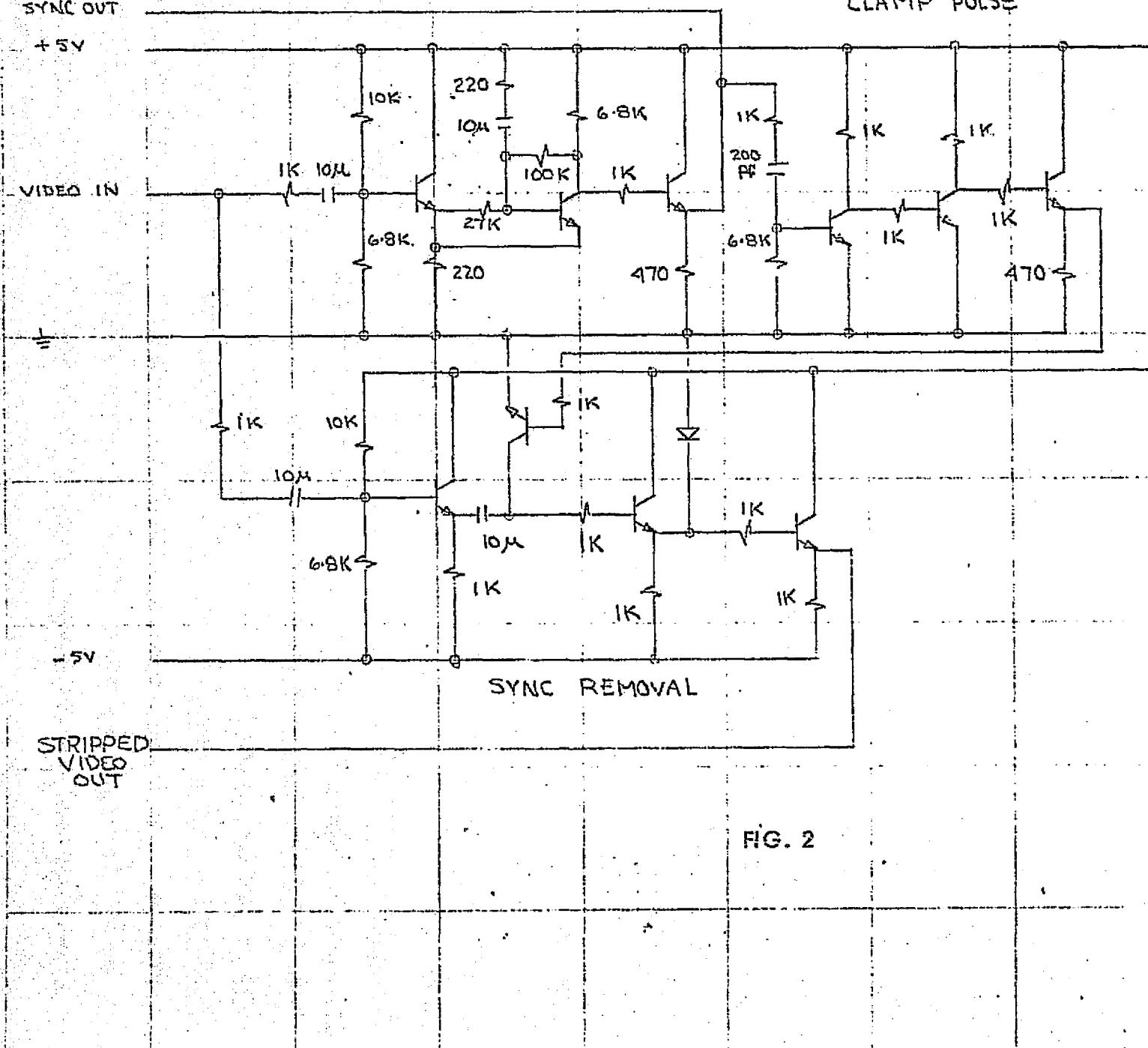
=

-5V

SYNC REMOVAL

STRIPPED
VIDEO
OUT

FIG. 2



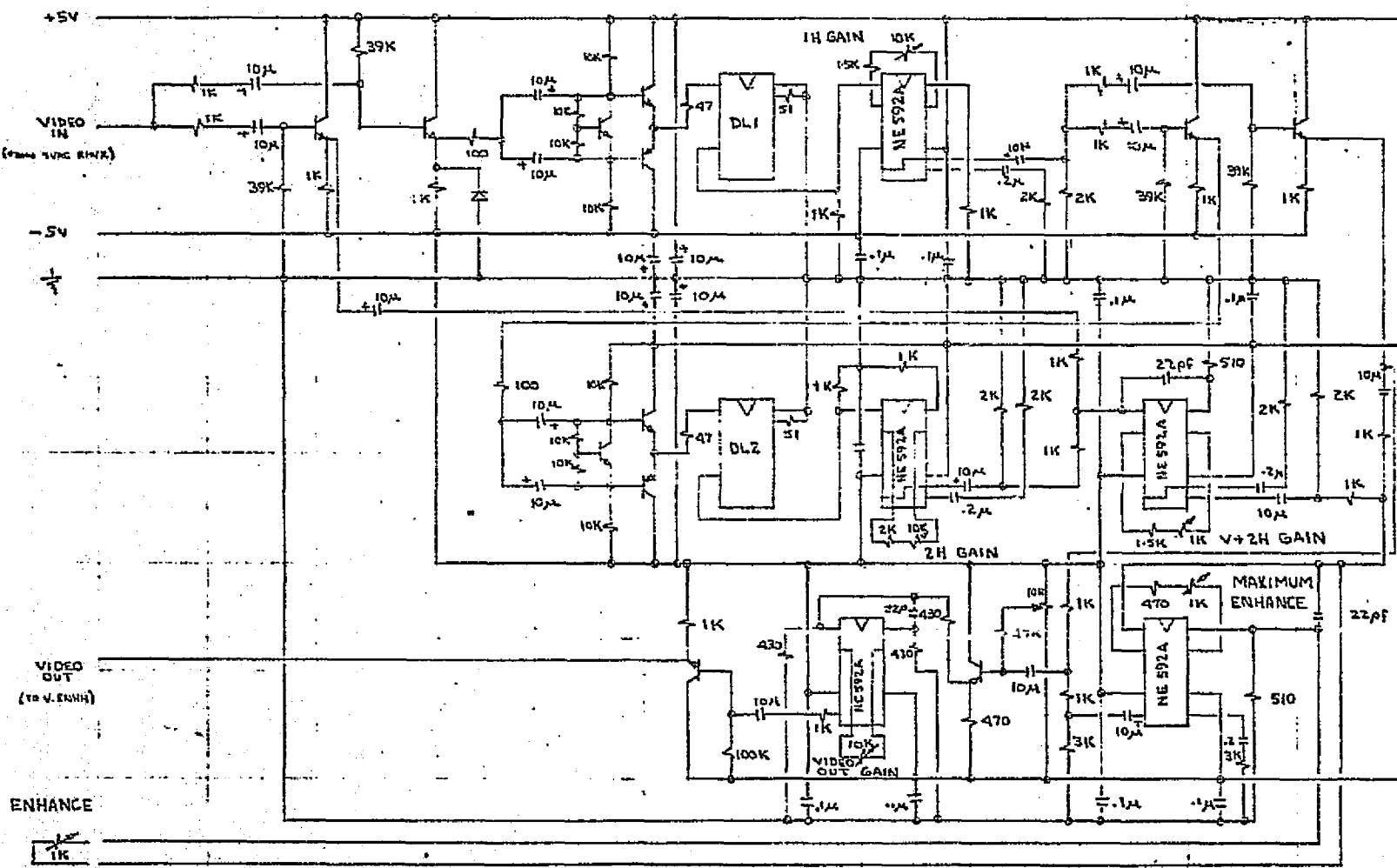
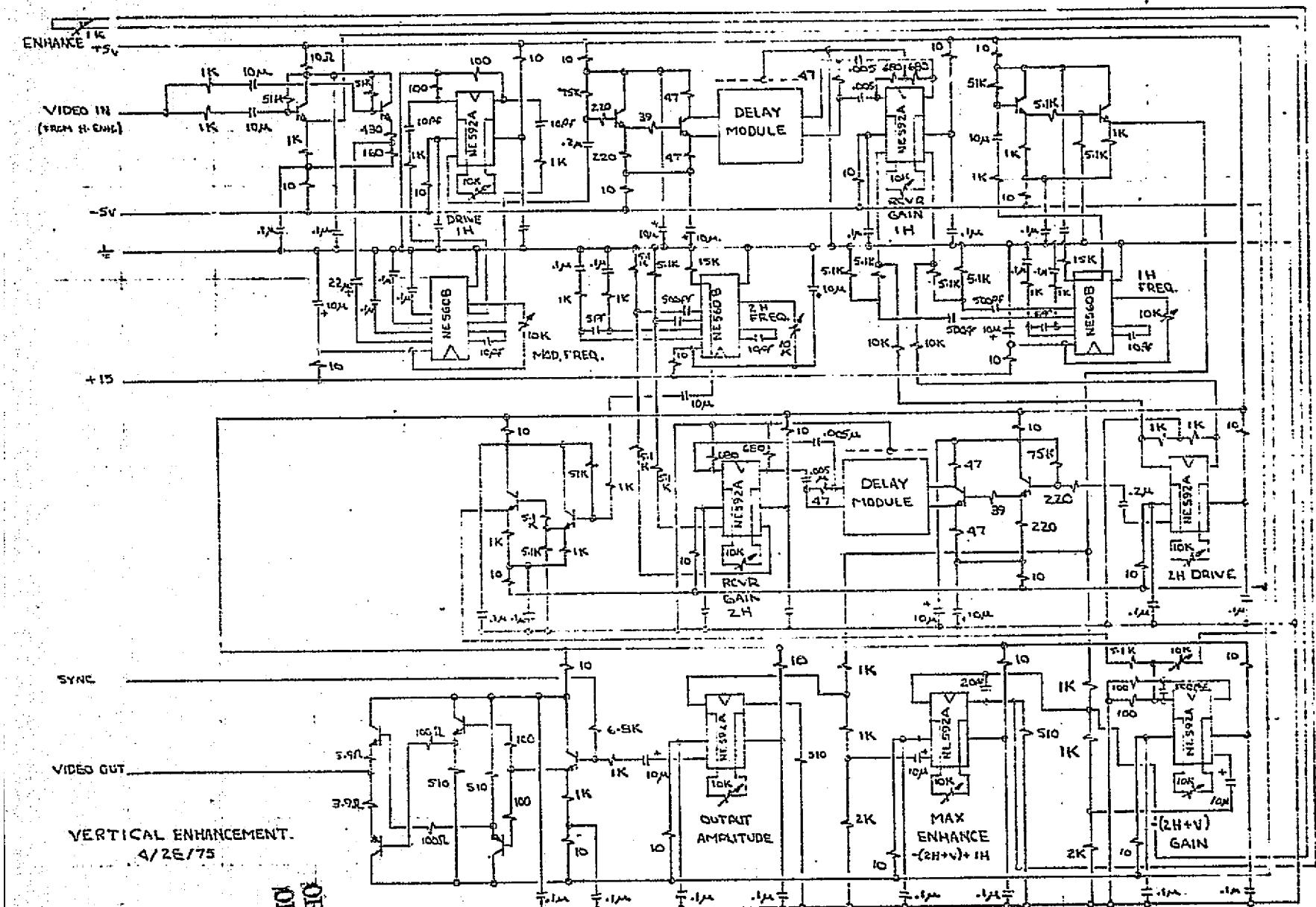


FIG. 3



ALL TRANSISTORS -
2N4124
ALL TRIM-POTS -
BOURNS 3389 P
DELAY MODULES -
CORNING 854227
63-492-A

FIG. 4

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VERTICAL ENHANCEMENT.
4/26/75

VIDEO OUT

SYNC

ENHANCE TS4



FIG. 5

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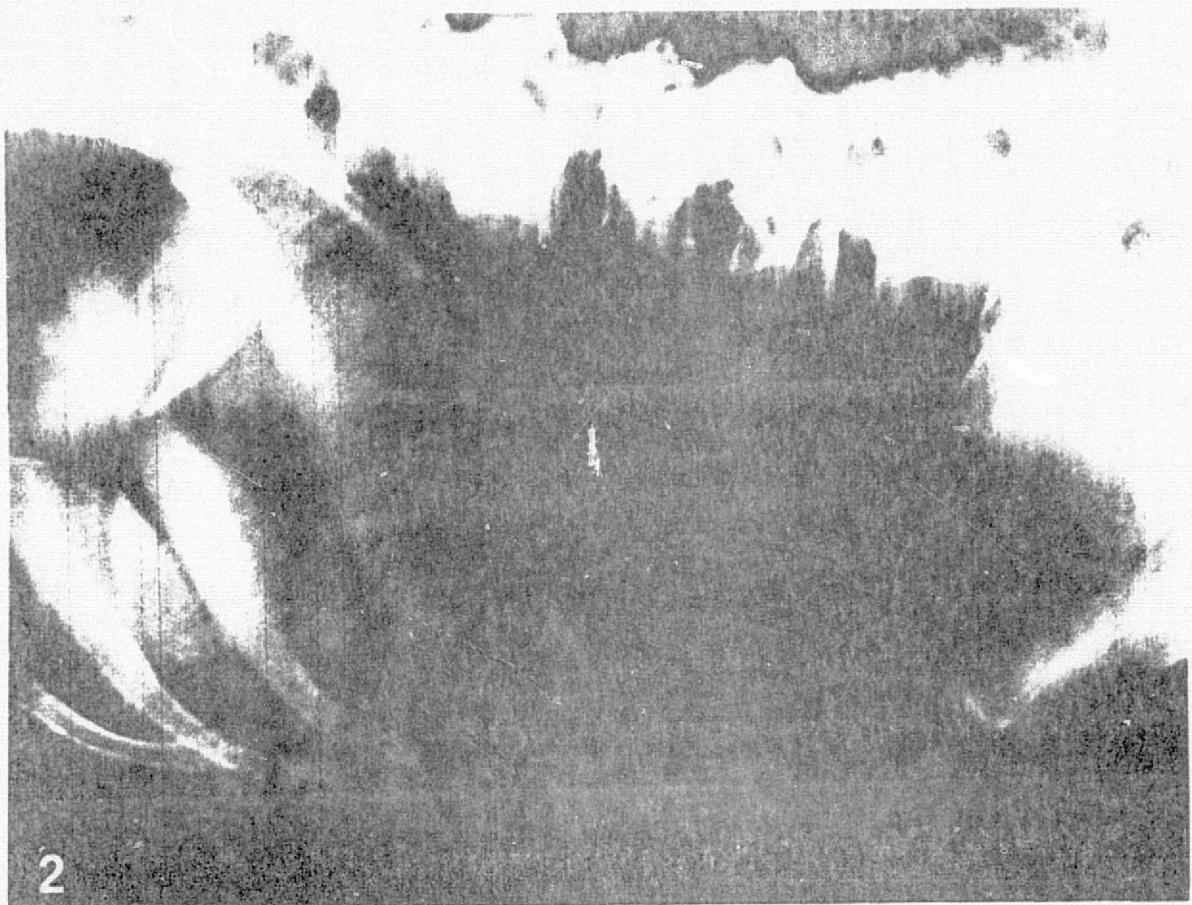


FIG. 6

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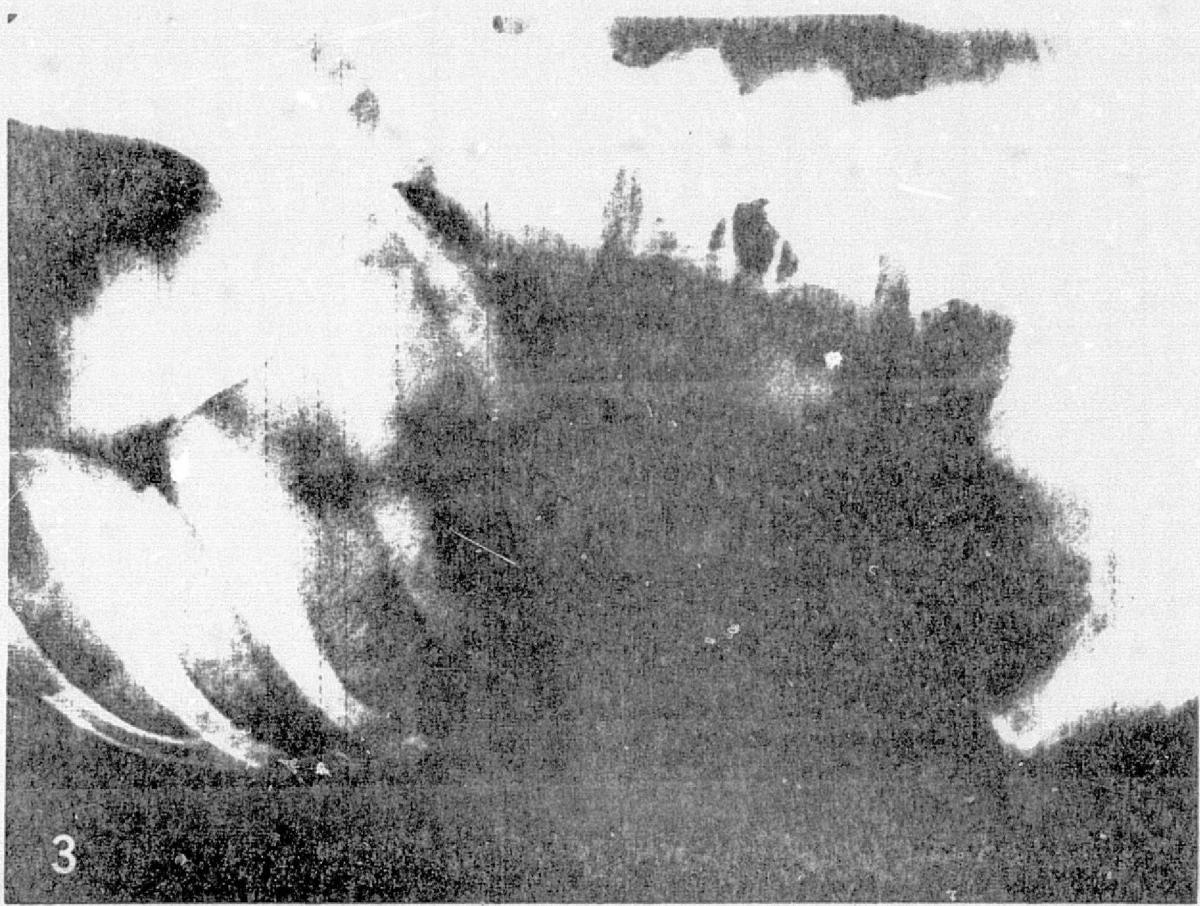


FIG. 7

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FIG. 8

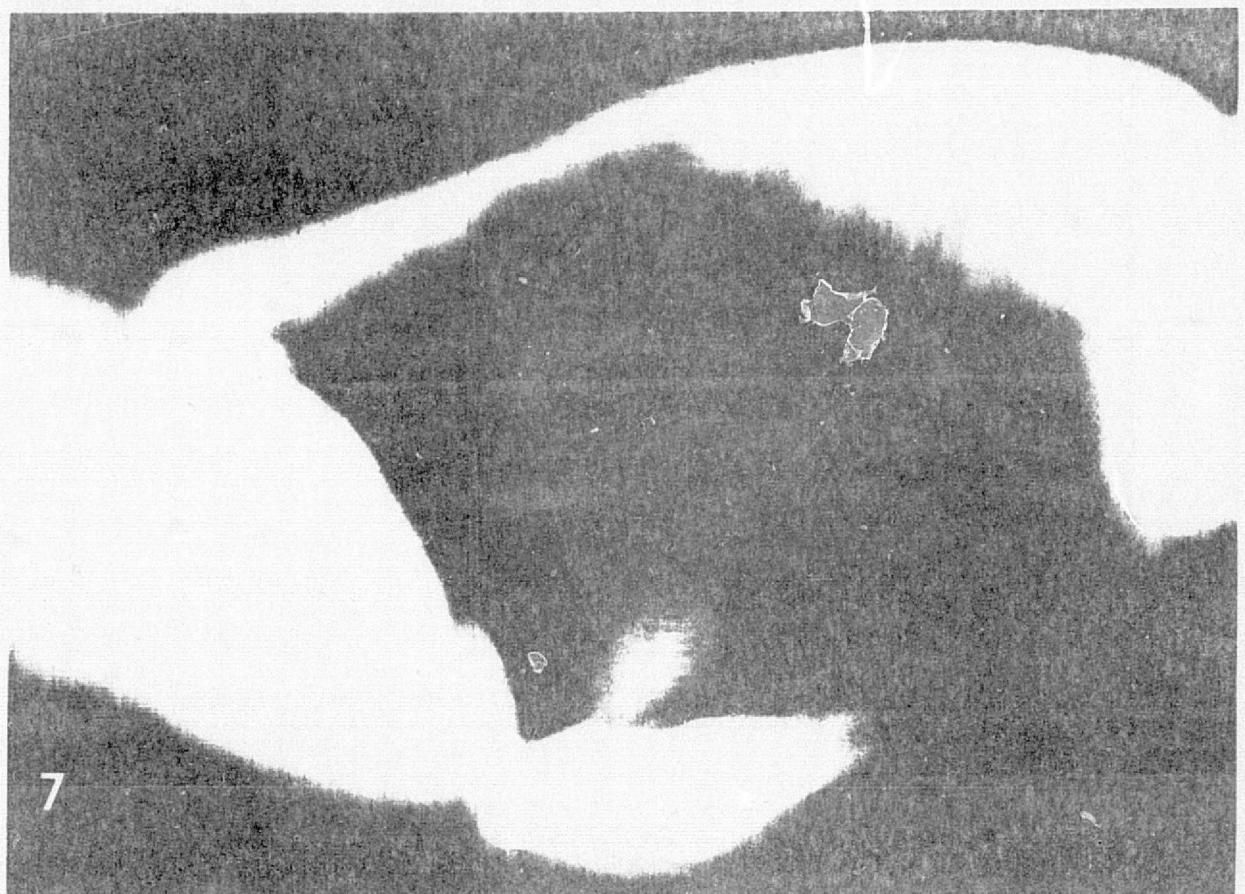


FIG. 9

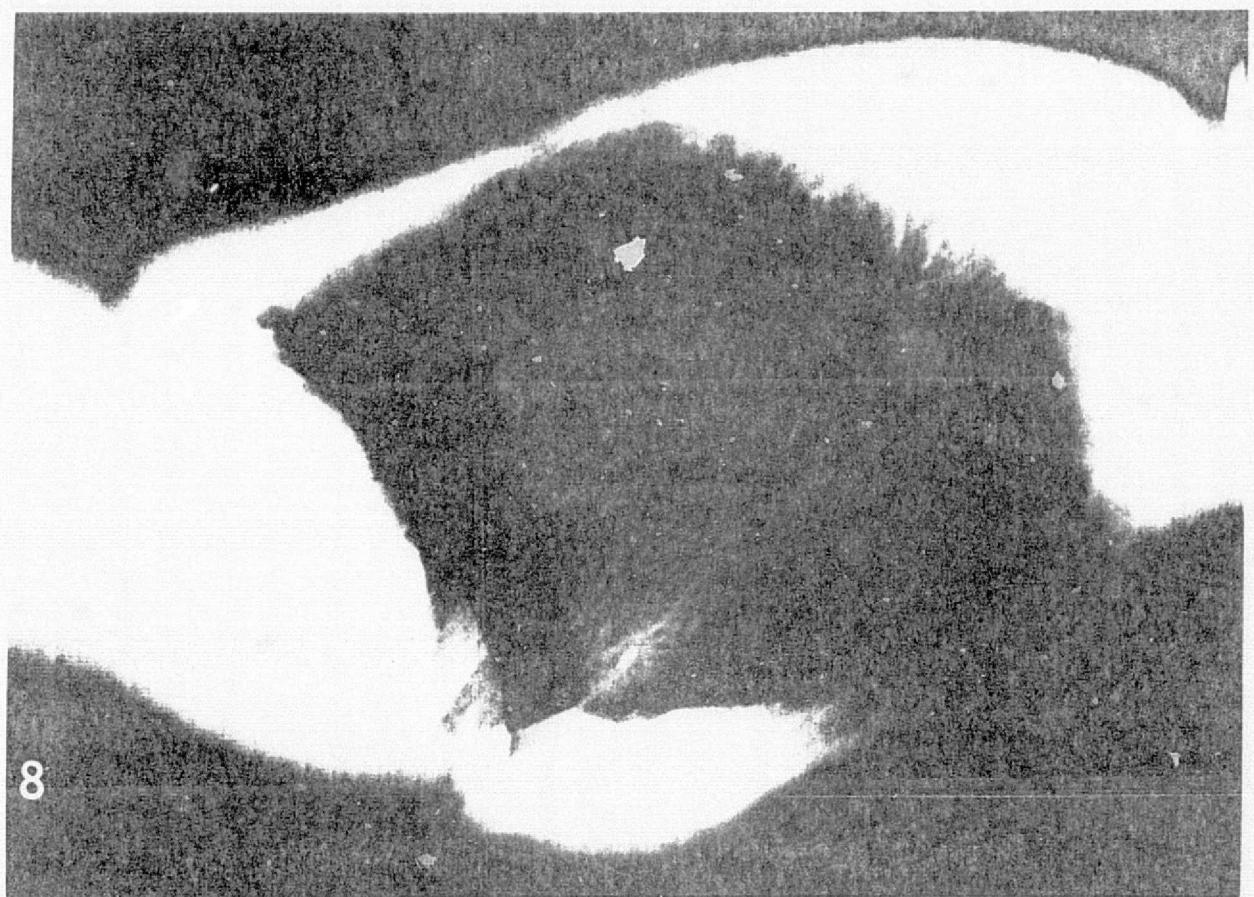


FIG. 10

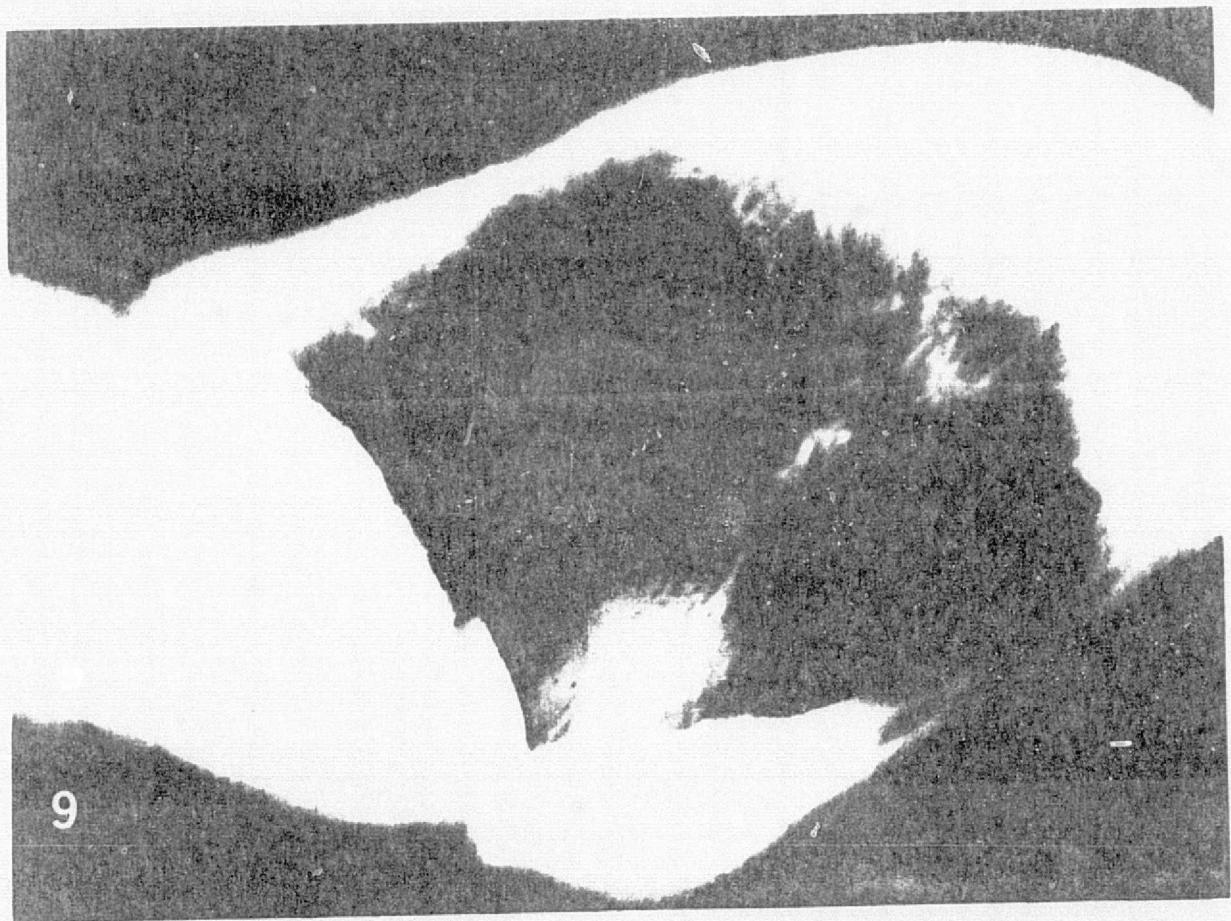


FIG. 11

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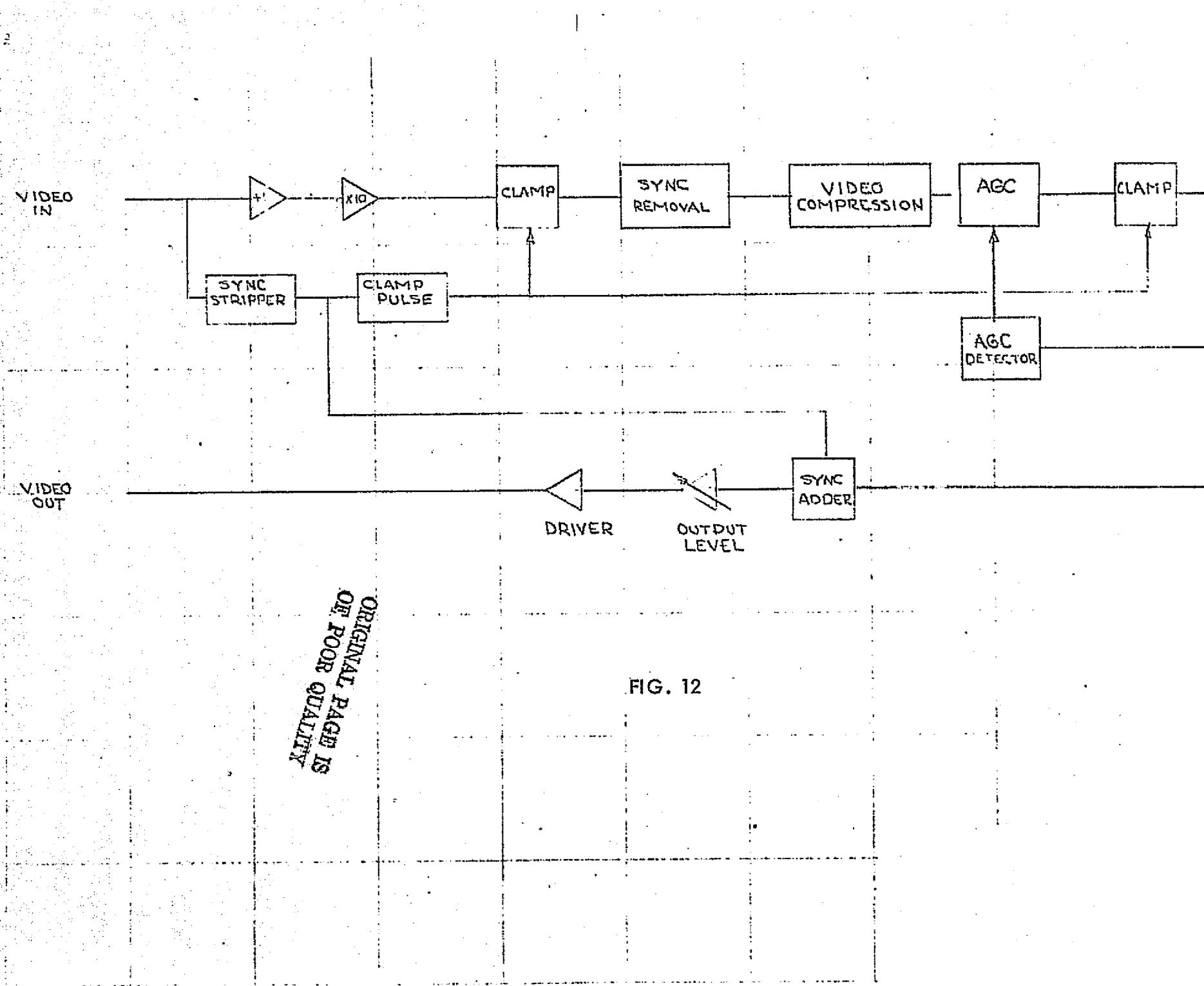


FIG. 12

NPN - 2N4124
 PNP - 2N4126
 FET - 2N5457
 DIODES - 1N914

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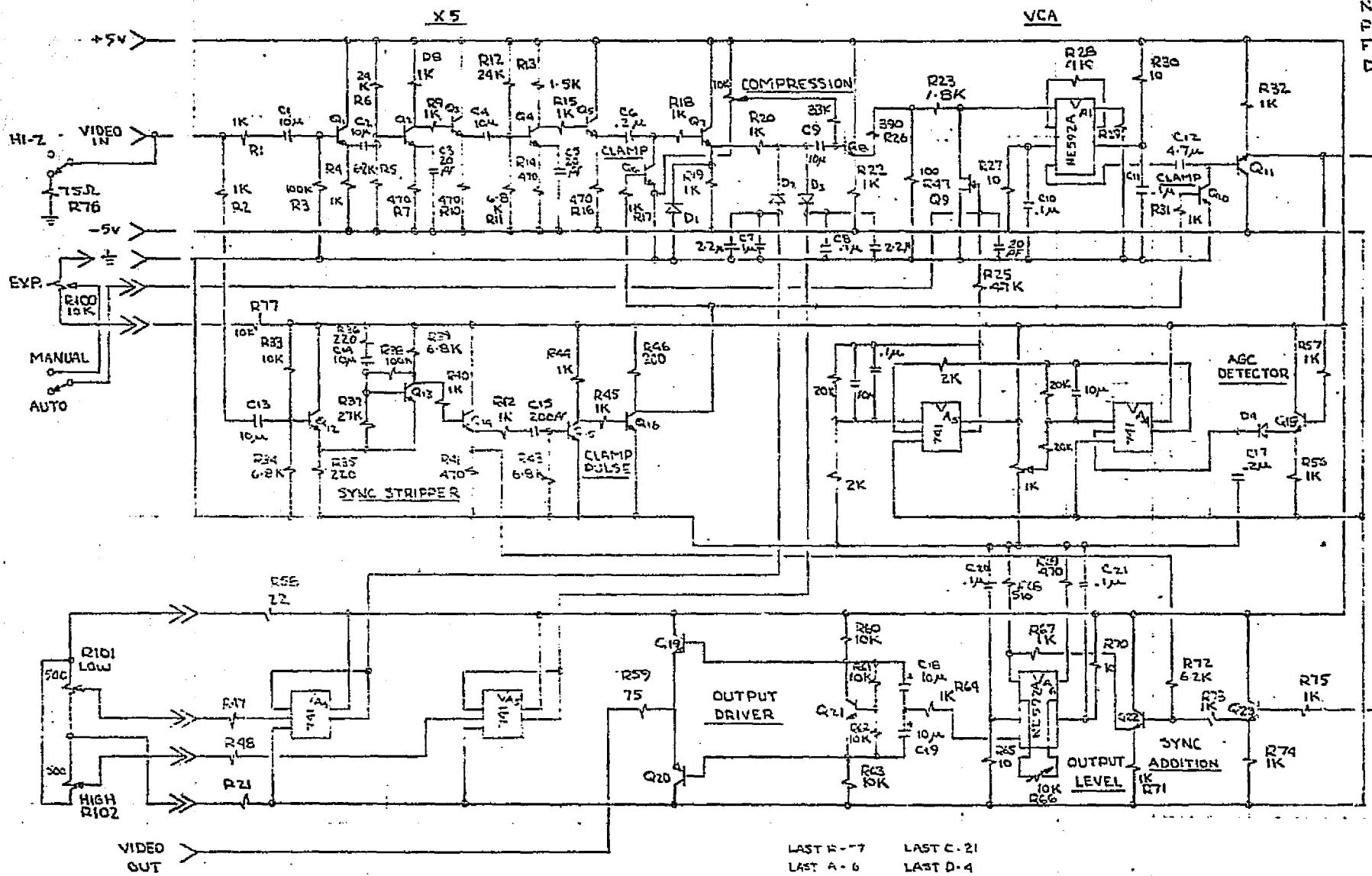


FIG. 13

Detail enhancement is achieved through adding of this detail signal to the main signal. The detail signal is produced by a subtraction and contains the edges and fine structures of the original image. The subtraction consists of generating low frequency (unsharp) images and adding these to the original. Vertical enhancement is achieved by delaying portions of a line by one and two lines ($64 \mu\text{ sec}$, $128 \mu\text{ sec}$) then adding one line to the next in the desired amount. The proportions can be varied from zero to approximately 50 percent. The horizontal enhancement is achieved similarly by two delay lines, whose values can be adjusted between 200 and 450 nanoseconds. Again, varying amounts of this detail signal may be mixed with the original video. The circuits for this device are given in Figs. 2, 3, and 4.

Examples of its performance are shown in Figs. 5-8. Fig. 5 is an unprocessed cholecystogram. Fig. 6 depicts the same image after passage through the four-way aperture equalizer. The enhancement of noise is quite evident. Fig. 7 shows the effect of a low-pass filter on this image. The effect is negligible. However, if the noise in the original image (Fig. 5) is reduced by the use of the sharp cut-off low-pass filter, described above, and then processed by the four-way aperture equalizer, considerable improvement is found. This is illustrated in Fig. 8. Another set of examples is shown in Figs. 9, 10, and 11. Fig. 9 is a roentgenogram showing the lateral view of a human chest. This is the original film. Fig. 10 shows this image after filtration and use of the four-way aperture equalizer. Fig. 11 shows what can be accomplished by the addition of a device described in (c) below and operated in such a manner as to produce compression of high video levels (whites) in the image while revamping those levels below this to the full video range.

(c) Selective Video Expander.

This device is capable of compressing either end of the video scale. That is either the low voltages (blacks) or the high voltages (whites) can be compressed in signal level from that which is generated by the x-ray beam. The balance of the signal is then expanded to occupy the full video signal range. Black or white compression can be used either singly or together and the degree of compression is selectable. A block diagram is shown in Fig. 12 and a circuit drawing in Fig. 13. The resultant image then has an



FIG. 14



FIG. 15

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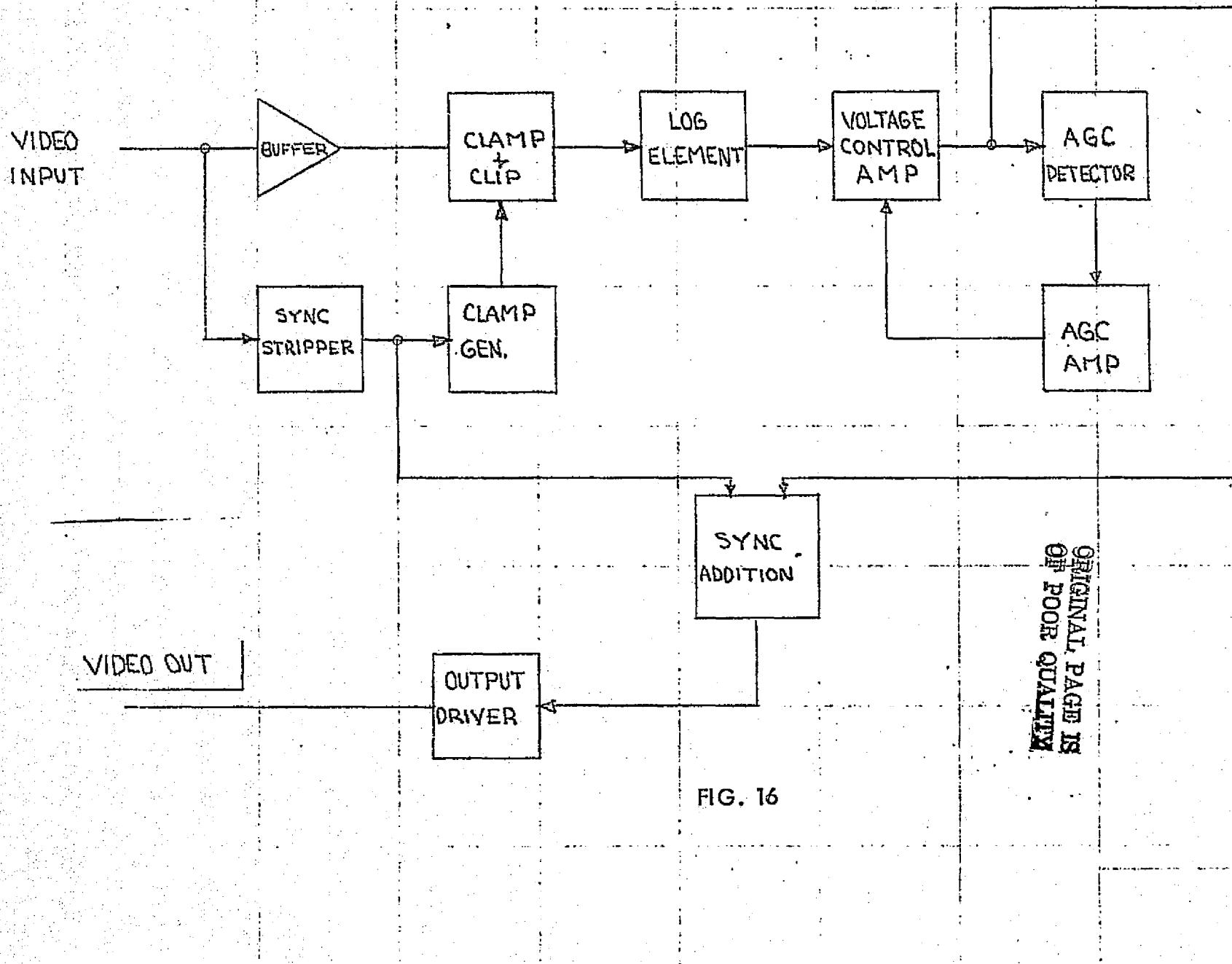
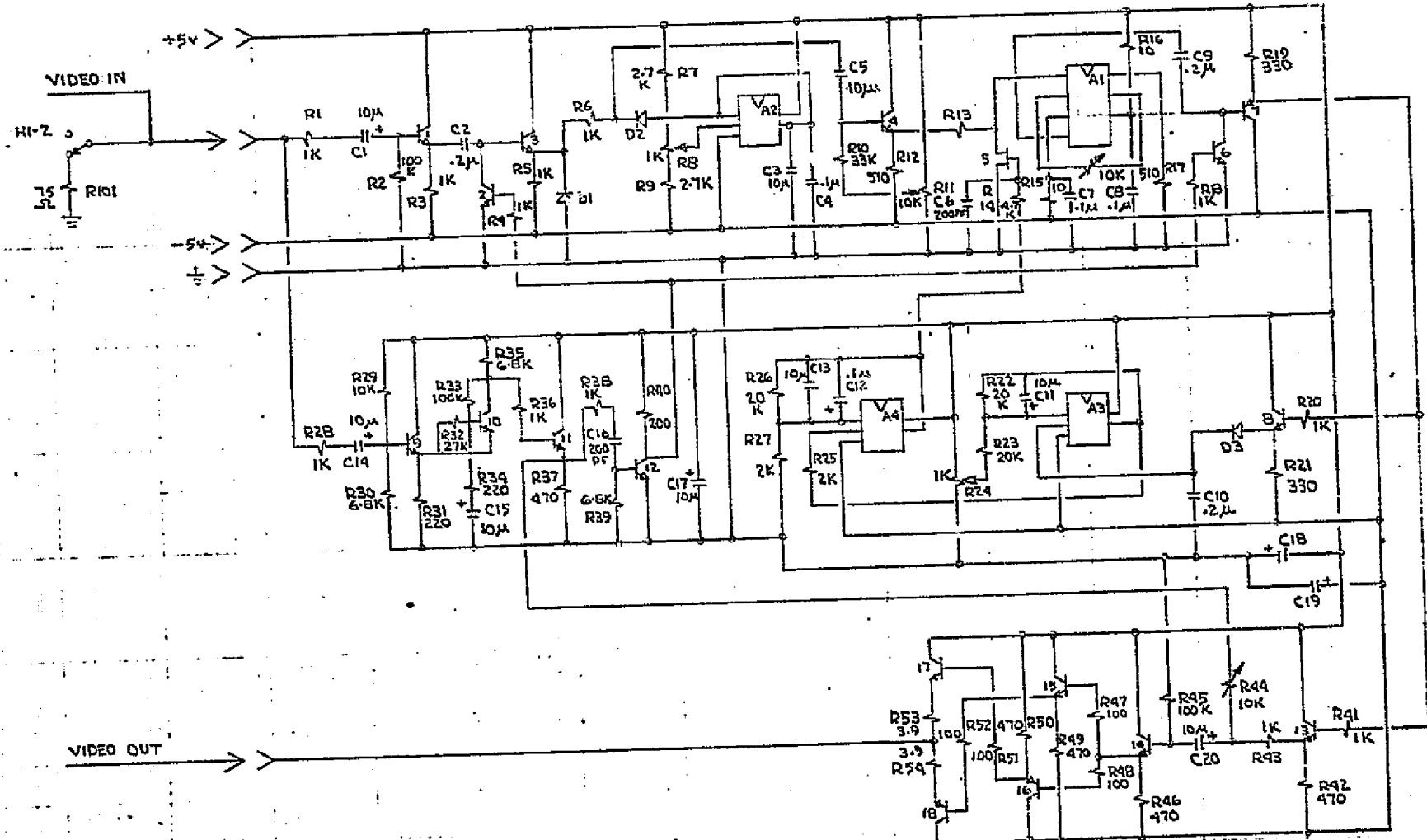


FIG. 16



R:54, C:29, Q:1B, A:4, D:3

FIG. 17

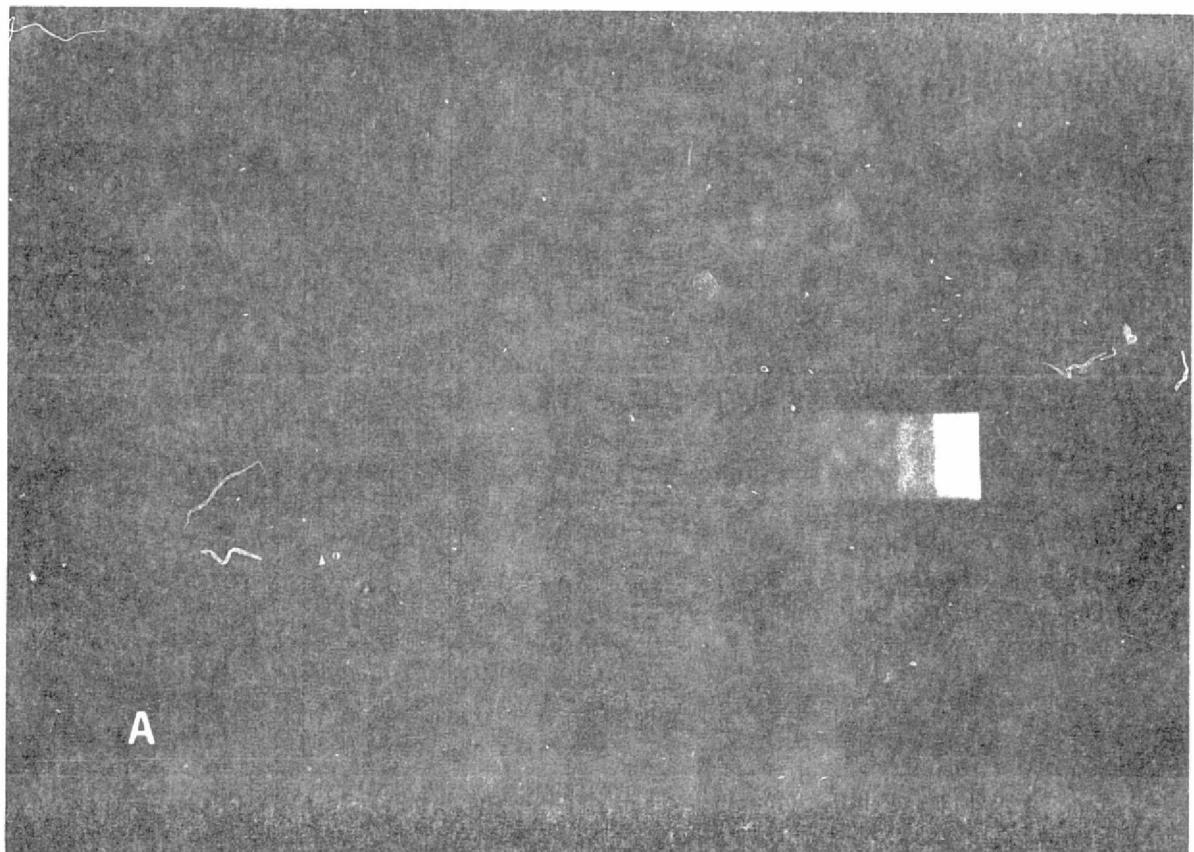


FIG. 18

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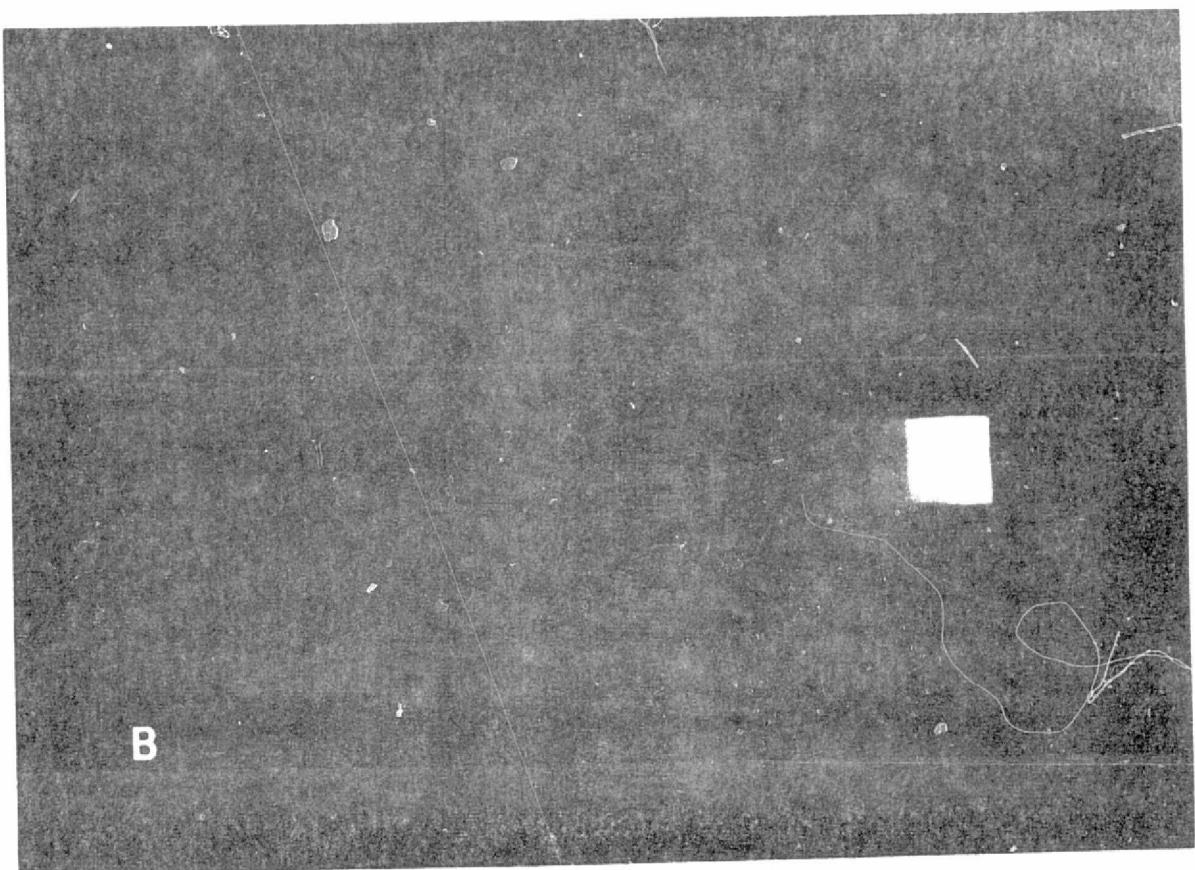


FIG. 19

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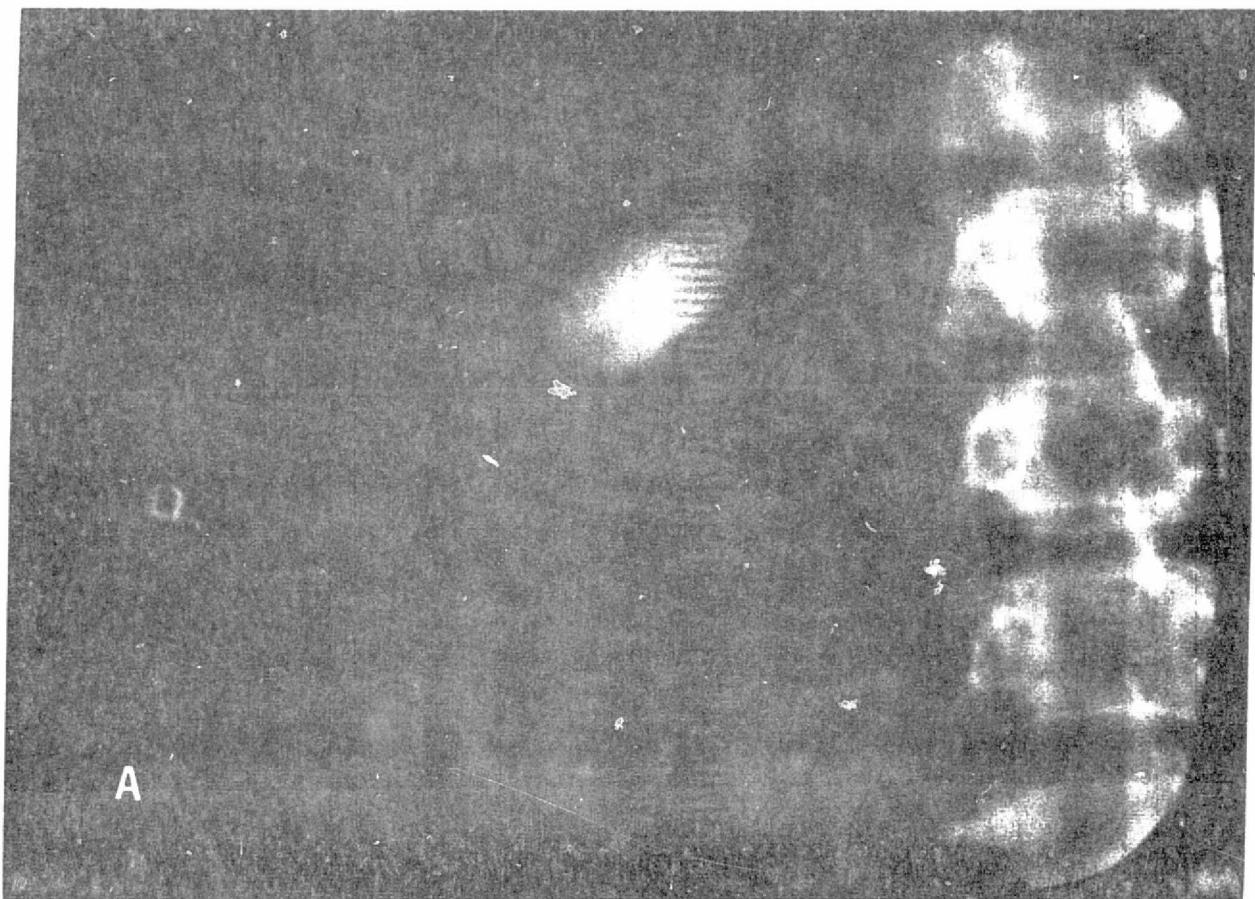


FIG. 20

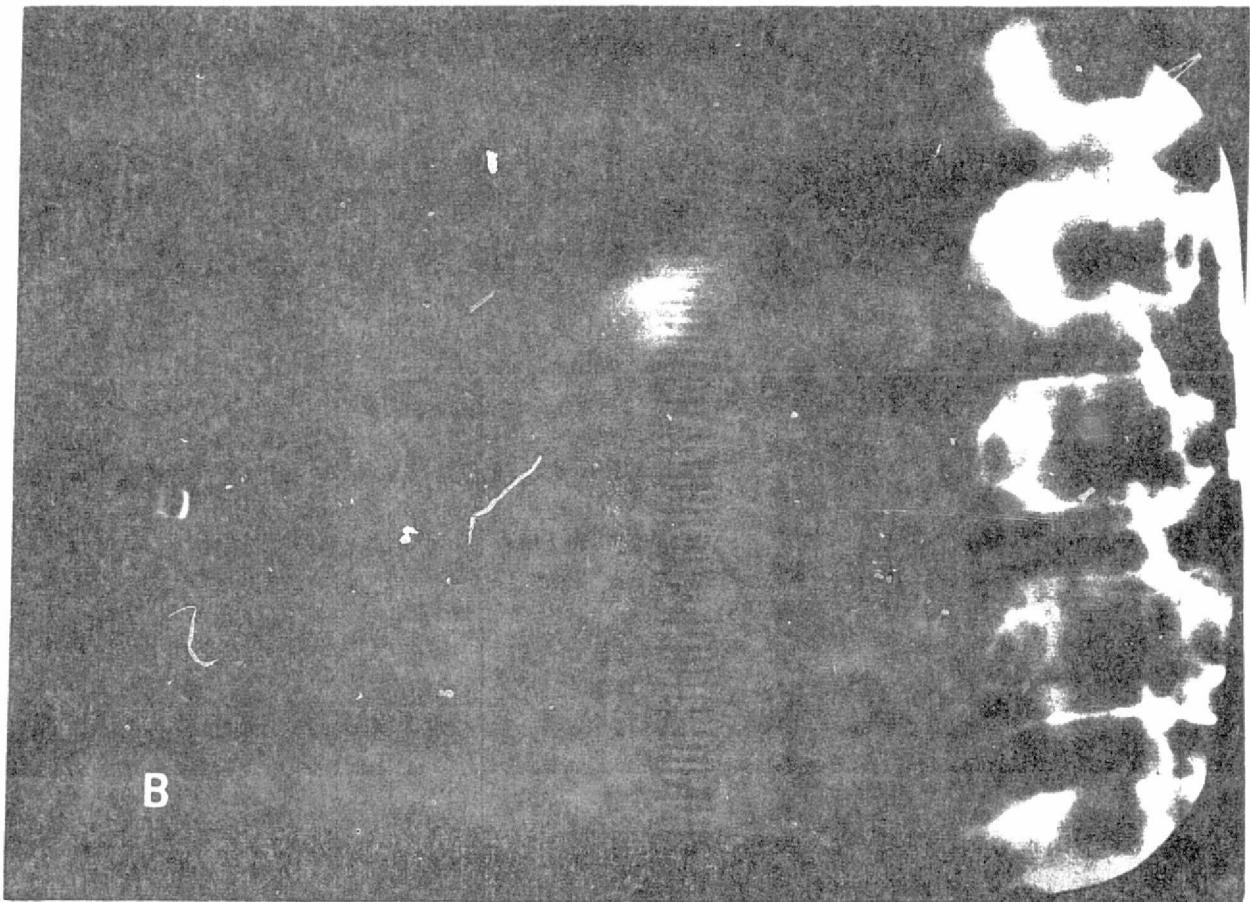


FIG. 21

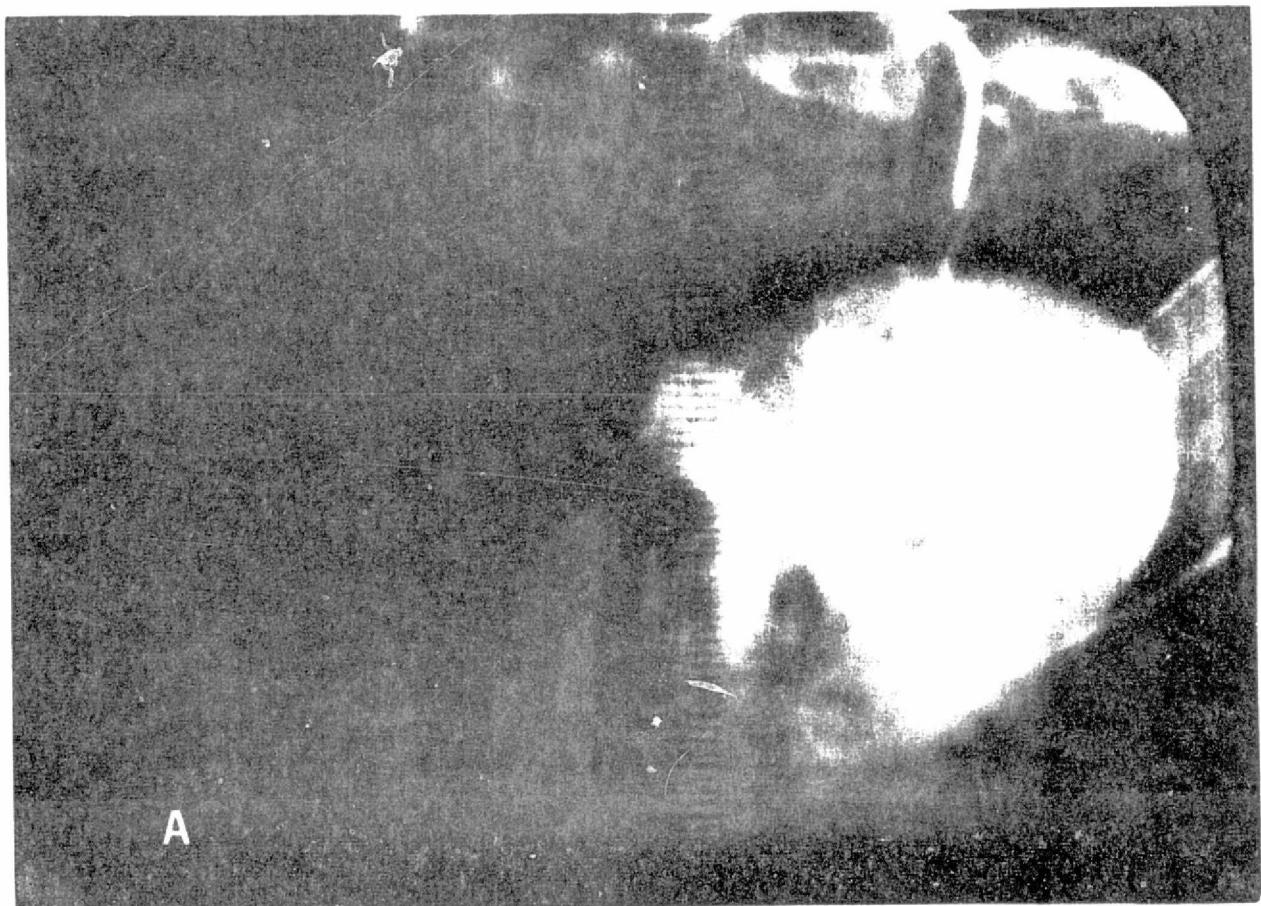


FIG. 22

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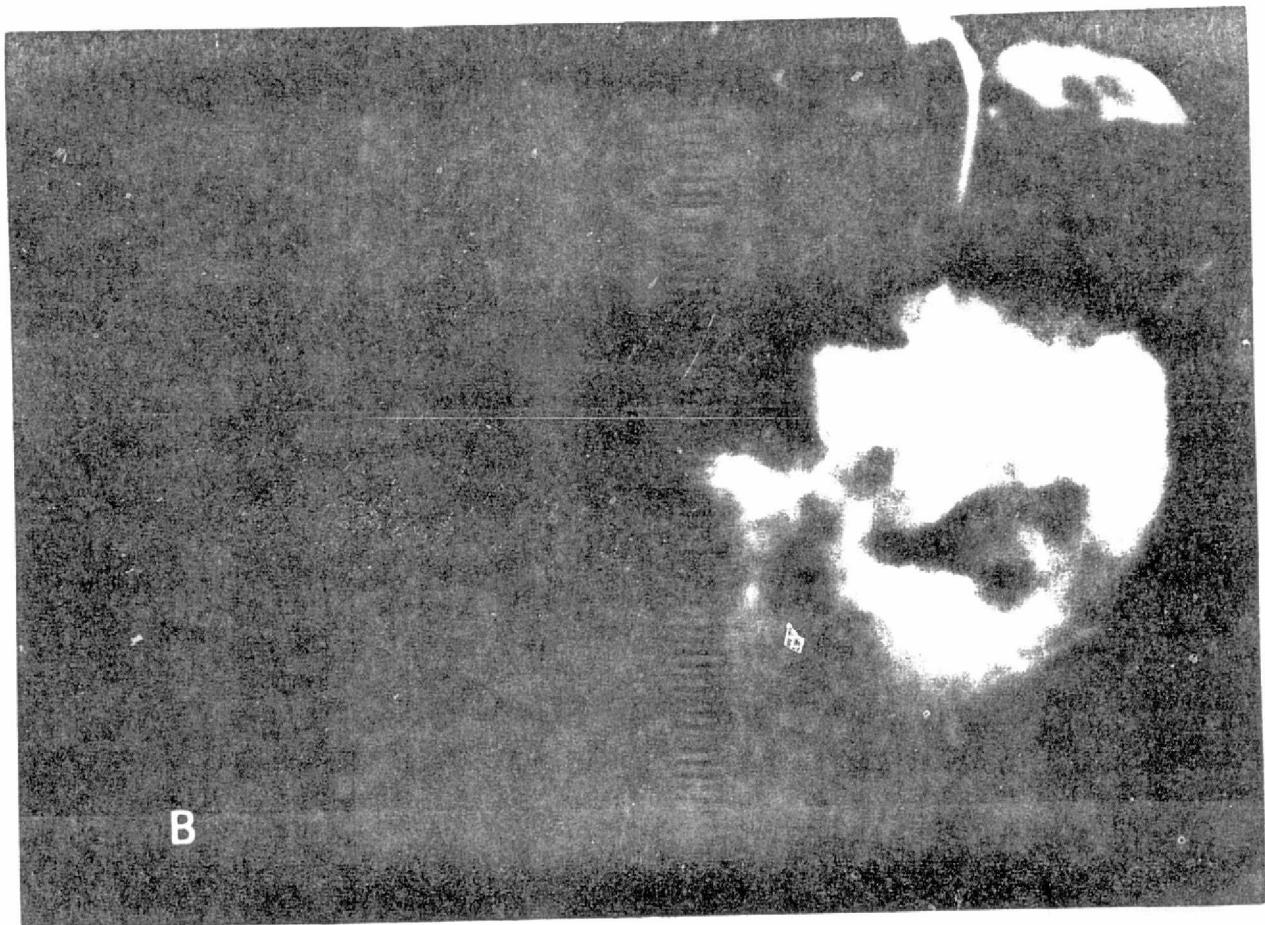


FIG. 23

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expanded gray scale making small contrast differentials readily visible. Fig. 14 shows the same cholecystogram shown in Fig. 8 processed to compress the whites. Fig. 15 shows this same image with black compression instead of the white compression whose effect was shown in Fig. 14.

(d) Exponential processor.

Another type of device which is useful for the expansion of contrast differentials in images and which will be useful either for visual display or computer analysis is an amplifier that has an output which is exponential with respect to its input. Having variable gain control allows selection of the proper portion of the video signal to be matched to the proper portion of the exponential function so that almost any portion of the video signal can be selected for enhancement.

The block diagram of this device is shown in Fig. 16 and the circuit in Fig. 17. The effect on the roentgenographic image of a step wedge is shown in Figs. 18 and 19. Fig. 18 is the original image while Fig. 19 is that after processing by the selective exponential amplifier. Two typical applications are shown in Figs. 20-23. Figs. 20 and 21 show a cholecystogram before and after processing respectively. In Figs. 22 and 23 we see this processor applied to a selective arteriogram of the left kidney. Fig. 22 is without processing, while Fig. 23 has been enhanced by the use of this circuit.

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1. A Simple Method for the Generation of Organ and Vessel Contours from Roentgenographic or Fluoroscopic Images, J.D. Newell, R.A. Keller, and N.A. Baily. *Medical Physics* 2:73-75 (1975).
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Computerized Tomography Using Video Recorded Fluoroscopic Images, A.C. Kak, C.V. Jakowatz, Jr., N.A. Baily, and R.A. Keller. Topical Meeting on Image Processing for 2-D and 3-D Reconstruction from Projections: Theory and Practice in Medicine and the Physical Sciences, Stanford University, July 10-12, 1975.

PAPERS IN PRESS

1. The Relationship of Video Response to X-Ray Input of Fluoroscopic Systems, N.A. Baily and R.A. Keller, Investigative Radiology.
2. A Computer Program for the Localization of Small Areas in Roentgenological Images, R.A. Keller and N.A. Baily, Investigative Radiology.
3. Acquisition of Quantitative Physiological Data and Computerized Image Reconstruction Using a Single Scan TV System, N.A. Baily. Proc. of the Conf. on Cardiovascular Imaging and Image Processing: Ultrasound, Angiography, and Isotopes. Stanford University.
4. Computerized Tomography Using Video Recorded Fluoroscopic Images, A.C. Kak, C.V. Jakowatz, Jr., N.A. Baily, and R.A. Keller. Proc. of the Topical Meeting on Image Processing for 2-D and 3-D Reconstruction from Projections: Theory and Practice in Medicine and the Physical Sciences.
5. The Response of Fluoroscopic Image Intensifier-TV Systems, N.A. Baily and R.A. Keller. Investigative Radiology.

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2. The Capability of Fluoroscopic Systems for the Production of Computerized Axial Tomograms, N.A. Baily, R.A. Keller, C.V. Jakowitz, Jr., and A.C. Kak.
To Investigative Radiology.

PERSONNEL PARTICIPATING IN PROGRAM

1. Norman A. Baily, Professor of Radiology
2. Robert A. Keller, Associate Specialist, Radiology
3. Earl M. Raeburn, Lab Technician
4. Richard J. Nachazel, Lab Technician

APPENDIX A

THE RESPONSE OF FLUOROSCOPIC IMAGE INTENSIFIER-TV SYSTEMS

Norman A. Baily, Ph.D. and Robert A. Keller, Ph.D.

ABSTRACT

Several recent articles have appeared in the literature which have assumed that the video signal output from a fluoroscopic x-ray image intensifier-TV chain is linear with the x-ray intensity presented to the fluoroscopic screen.^{4,7,8} Silverman, and Baily and Crepeau have pointed out that contrast as measured by the video signal level is linear with absorber thickness placed in the beam. The assumption of linearity would not allow for this experimental result. We have, therefore, investigated the response from two types of x-ray image intensifiers and three types of fluoroscopic TV chains.

The response of these systems was studied by digitization of uniformly irradiated areas recorded on video disc. Beam intensity was monitored through the use of a transmission ionization chamber and varied by inserting various thicknesses of aluminum in the path of the beam. To simulate typical fluoroscopic conditions the beam traversed 15 cm of water before incidence on the aluminum absorbers.

Our results indicate that the video signal output is linear with absorber thickness over much of the useful operating range of the fluoroscopic systems. This is not consistent with the assumption of linearity with input intensity nor with direct application of the Lambert-Beers law to the amplitudes of video signals.

Our results indicate that video signal amplitude or gray level response is linear in μd , where μ is the effective absorption coefficient for the pertinent operating conditions and d is the thickness of the material traversed. The use of a fluoroscopic x-ray intensifier-TV chain for quantitative applications requires careful calibration.

KEY WORDS: attenuation; digitization; fluoroscopy; gray scale; television.

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La Jolla, California 92093.

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Recently several articles have appeared in both the radiological and computer literature in which video signal levels were utilized for quantitative purposes and in which it was tacitly assumed that this signal level was linearly related to the x-ray beam intensity incident on the input screen. Indeed this impression is found among many radiologists, radiological physicists, and x-ray engineering personnel. At least two well known factors make this type of response very unlikely. First, the camera tubes have, in general, a power law response to input light. That is, the video (electrical) signal out is proportional to some power (usually less than one) of the input light intensity. Second, the light out of x-ray fluorescent materials is proportional to the energy absorbed by this material. Since this quantity is wavelength dependent and the beam is heterogeneous, the response will depend on the thickness of the absorber being imaged. This is caused by the change in beam quality produced by the absorber and the presence of scattered radiation. In addition, x-ray image intensifiers, light amplifiers, etc. also have response functions which can be non-linear with input intensity.

We have found ³ in our investigations of fluoroscopic TV chains that the contrast between two adjacent areas is a linear function of the absorber thickness over a large portion of the system's response. Again, this could not be the case if signal output were linearly related to x-ray intensity at the input surface since this quantity should be approximately exponentially related to absorber thickness. Another experimental finding, related to computerized reconstruction of transverse body sections from video disc recorded fluoroscopic images, showed that images could not be formed when that assumption was made except for situations embodying large degrees of contrast but that rather excellent images can be reconstructed when a linear relationship between the product of attenuation coefficient and absorber thickness is used in the reconstruction algorithm. We have, therefore, examined the response of a number of fluoroscopic-TV systems and have not found any that respond linearly with intensity. While not all systems have proven to be identical in response, their response is more likely to be closer to having a linear response with absorber thickness than to be linear with x-ray intensity incident on the input surface.

METHOD

Three different types of fluoroscopic TV chains were used in this investigation. Two of these were standard clinical units, one with a vidicon camera tube, the other with a plumbicon camera tube. The third unit is a large, flat-screen unit designed and built in our laboratory. This is similar in configuration to that described by Baily and Crepeau² except that it has a 14" x 17" input screen, a 1.5" vidicon, and a 40 mm three-stage light amplifier. In each, an x-ray beam generated at 100 kVp and having a half value layer in excess of 3 mm Al was used. To make the situation somewhat comparable to that which would exist clinically, this beam was passed through 10 cm of H_2O before aluminum absorbers of varying thickness were introduced. The initial beam intensity (tube output) was monitored by using a large area, thin window (0.00025" aluminized mylar), parallel plate ionization chamber. The beam after passing through the absorber material fell directly on the input surface. The video image was then recorded on a video disc. Five recordings were made at each absorber thickness.

The video image was digitized directly from the disc recording and quantized into 128 gray levels.¹ That is, the full amplitude of the video signal (from black to white), which is usually about 0.7 volts, was divided into 128 discrete bins. A particular response to a given roentgen input then was assigned to one of these bins.

The five recordings were then averaged on a point-to-point basis. The central 900 averaged points were again averaged to yield a value for the gray level assigned to a particular image. This 30 by 30 matrix of points corresponds to input screen areas of 29 cm^2 , 8.2 cm^2 , and 3.6 cm^2 for the three units investigated. In addition to these gray scale determinations, the quality of the incident beam (at the screen) was also investigated.

RESULTS

The data obtained using a G.E. Fluorocon having a vidicon camera tube in its TV chain is shown in Fig. 1. The range of gray levels lying between 107 and 22 represents the entire usable signal range of the camera's output. The break in the data at gray level 85 is probably due to the camera target voltage setting used in this particular unit. We have found that such irregularities can usually be eliminated by careful adjustment of

the camera parameter. The dynamic range can also be extended by lowering the slope of the camera response curve. However, this data was obtained from a unit which was in routine use and considered to be satisfactory for clinical use.

Fig. 2 shows the data obtained using our large area, flat screen imaging system. This system utilizes a vidicon camera tube optically coupled to a three-stage light amplifier. The gray scale response is linear from gray levels 102 to approximately 47. This is in excess of 50 percent of the usable signal range.

Fig. 3 represents the data obtained using a Philips 6" intensifier with a plumbicon camera tube. This combination's response is non-linear with absorber thickness over its entire response range and it is easily demonstrated that it is not exponential in character.

DISCUSSION

The response curves obtained for the three systems mentioned above are typical of both our previous experience using a G.E. Fluorocan 6" - 9" system having a vidicon camera tube¹ and that reported by Silverman.⁵ In our experience using x-ray fluoroscopic systems, we have failed to find any which respond linearly with x-ray input intensity. The vidicon or plumbicon response curves show that the electronic signal is proportional to some power (usually about 0.6) of the light intensity incident on the target. This relationship has been shown to also hold for single-scan operation and also for the signal levels associated with video disc recording.¹ The manufacturers data sheet for the three-stage light amplifier also shows a similar response. Since it is generally assumed that the light output of a fluorescent screen, or an x-ray image intensifier incorporating such a screen, has a linear response with absorbed energy it is difficult then to see why one should expect linearity of the output signal with incident x-ray intensity in any equipment of this type.

In an effort to gain a more complete understanding of the mechanisms involved in the true system response as measured, we have also looked at the character or quality of the incident beam. In our case, the well filtered beam emerging from the collimator showed good homogeneity. That is, after traversing a thickness of about 6 mm of lucite, the attenuation curve was essentially exponential in character. However, after traversing 10 cm of water, which is somewhat analogous to the usual fluoroscope beam, this situation

no longer prevails. An attenuation curve determined using this same beam after passage through the water scatterer showed a non-exponential behavior over the range of 0 to 1" of aluminum. In fact, the effective energy of the beam changed from 51.5 keV to 77.5 keV, a factor of approximately 1.5, in going through this range of attenuator thickness. The curve was still changing slope at the same rate after introduction of 1" of aluminum as at its start. From this information on the beam quality and our other data, we were able to compute a curve of gray level response vs. energy absorbed by the screen. This curve is shown in Fig. 4. Although this curve is very similar to that of the input x-ray intensity vs. video level output, it is not identical, since, as is the case for all diagnostic quality beams, this beam was non-homogeneous in energy. The data used to calculate this curve is that which we obtained using the large area, flat screen system whose response is shown in Fig. 2. There is considerable non-linearity shown in the data shown in Fig. 4 which reflects the non-linearity with input intensity. However, since the ratio of energy deposited to beam intensity is wavelength dependent, it is easy to see why the electronic signal output level is not linear with the x-ray intensity incident on the screen when one considers the data shown in Fig. 4. Although we have not studied the effects of scatter, beam geometry, etc. in detail, we have noted that our experimental response curves have all been quite similar under many varied conditions. For example, we have used water phantoms of 5, 10, 15, and 20 cm thickness. We have used fields as small as 5 cm x 5 cm and as large as 20 cm x 20 cm. In addition, we have on occasion used these systems without grids, with a 6:1 moving grid, and with a 12:1 stationary grid. It is our qualitative impression that variations in these geometric factors did not significantly change the shape of the response curves. Increases of distance between absorber and scatterer and the input screen, also did not seem to influence the degree of linearity.

It is easy to see why attempts to use fluoroscopic images as inputs to computerized tomography algorithms have resulted in the necessity for use of contrast agents.⁴ The assumption of signal output linearity with x-ray intensity results in a compression of the resulting gray scale differentials. In our own laboratory, using the system where response is shown in Fig. 3 and making the approximation that gray scale was a linear function of the product of absorption coefficient and absorber thickness, we have been able to

reconstruct tomograms of a rat thorax and visualize the heart, sternum, spine, ribs, and soft tissues without the use of any contrast agents.⁶

Finally, we should point out an earlier finding,¹ and that is the two-segment curve which is almost always obtained when using low input levels. This is also reflected in the data shown in Fig. 1.

CONCLUSIONS

1. In general, x-ray fluoroscopic systems do not respond linearly to incident x-ray intensity.
2. In many cases, x-ray fluoroscopic system response is a linear function of the product of attenuation coefficient and absorber thickness over significant portions of their operating range.
3. For any applications involving quantitative measurement or evaluation of the degree of attenuation due to passage through an absorber, the system to be used must be investigated thoroughly and its response function under the various operating conditions utilized be completely determined.

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FIGURE CAPTIONS

FIG. 1: Gray scale (digitized) response of a typical fluoroscopic-TV chain using a vidicon camera tube as a function of the absorber thickness placed in the incident x-ray beam.

FIG. 2: Gray scale (digitized) response of a large area (14" x 17"), flat screen fluoroscopic system utilizing a 1.5" vidicon camera tube as a function of absorber thickness.

FIG. 3: Gray scale (digitized) response of a fluoroscopic-TV chain using a plumbicon camera tube as a function of absorber thickness.

FIG. 4: Gray scale response as a function of energy absorbed by the fluorescent screen for the fluoroscopic system whose response is shown in Fig. 2.

FIG. 1

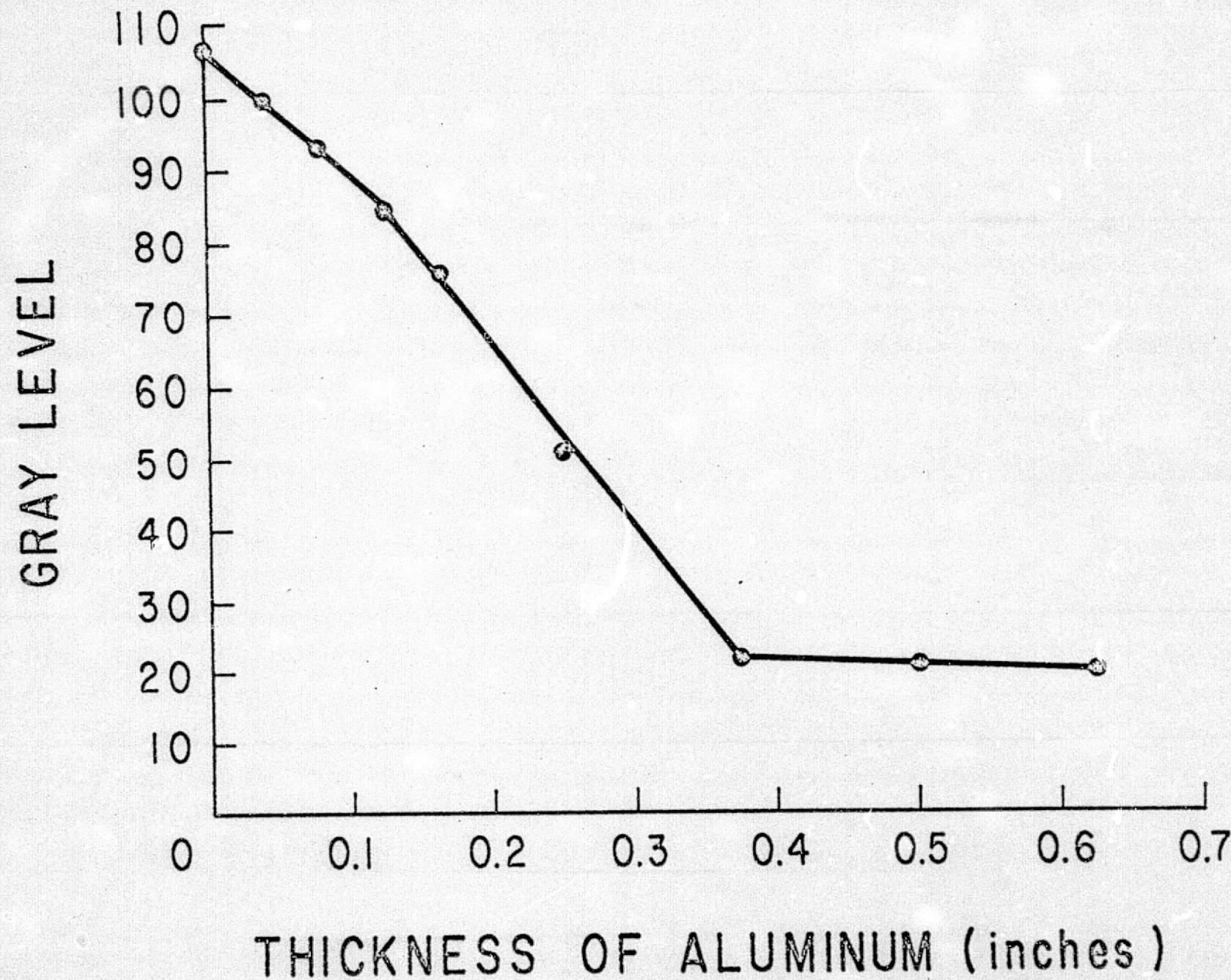


FIG. 2

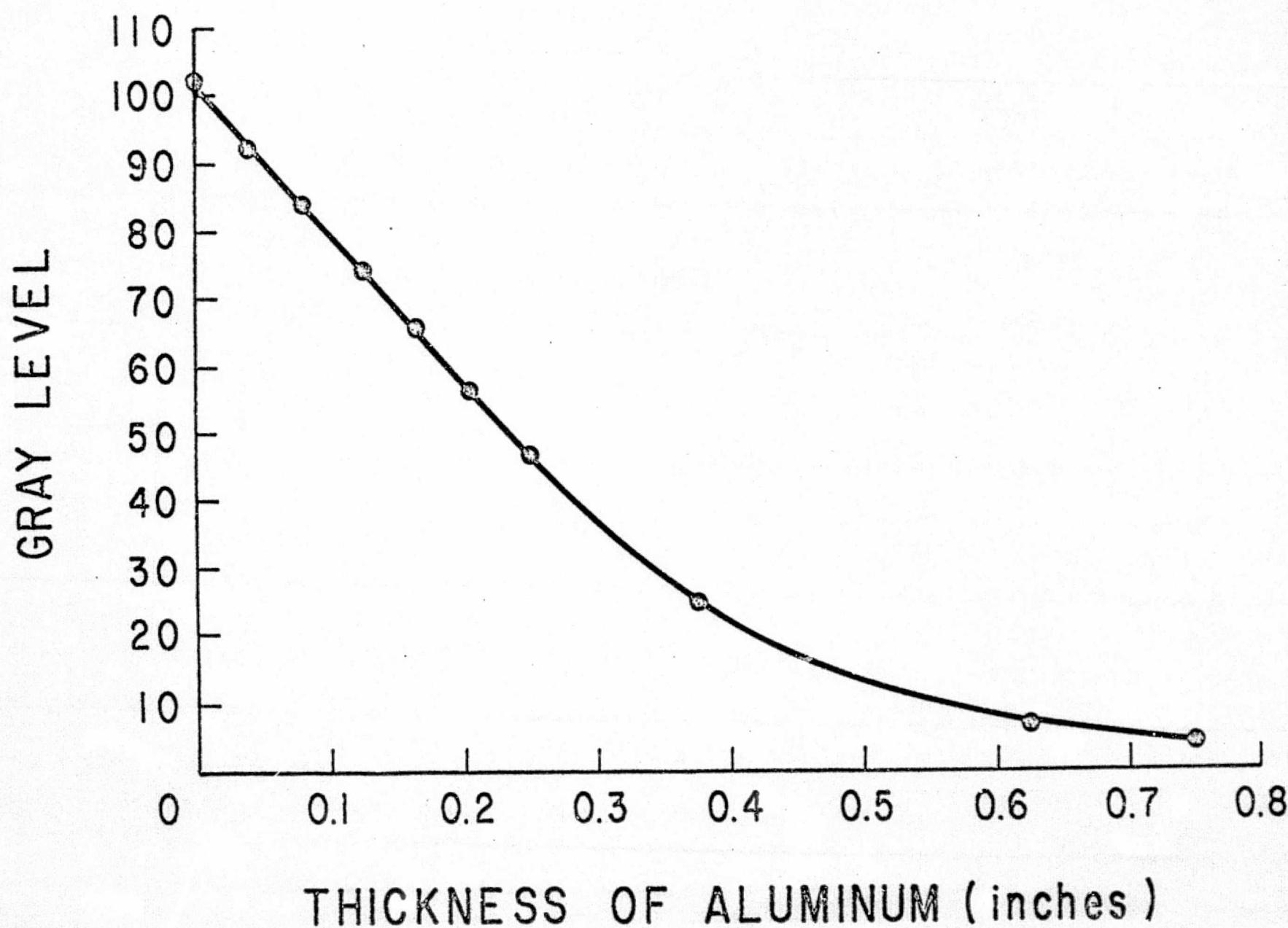


FIG. 3

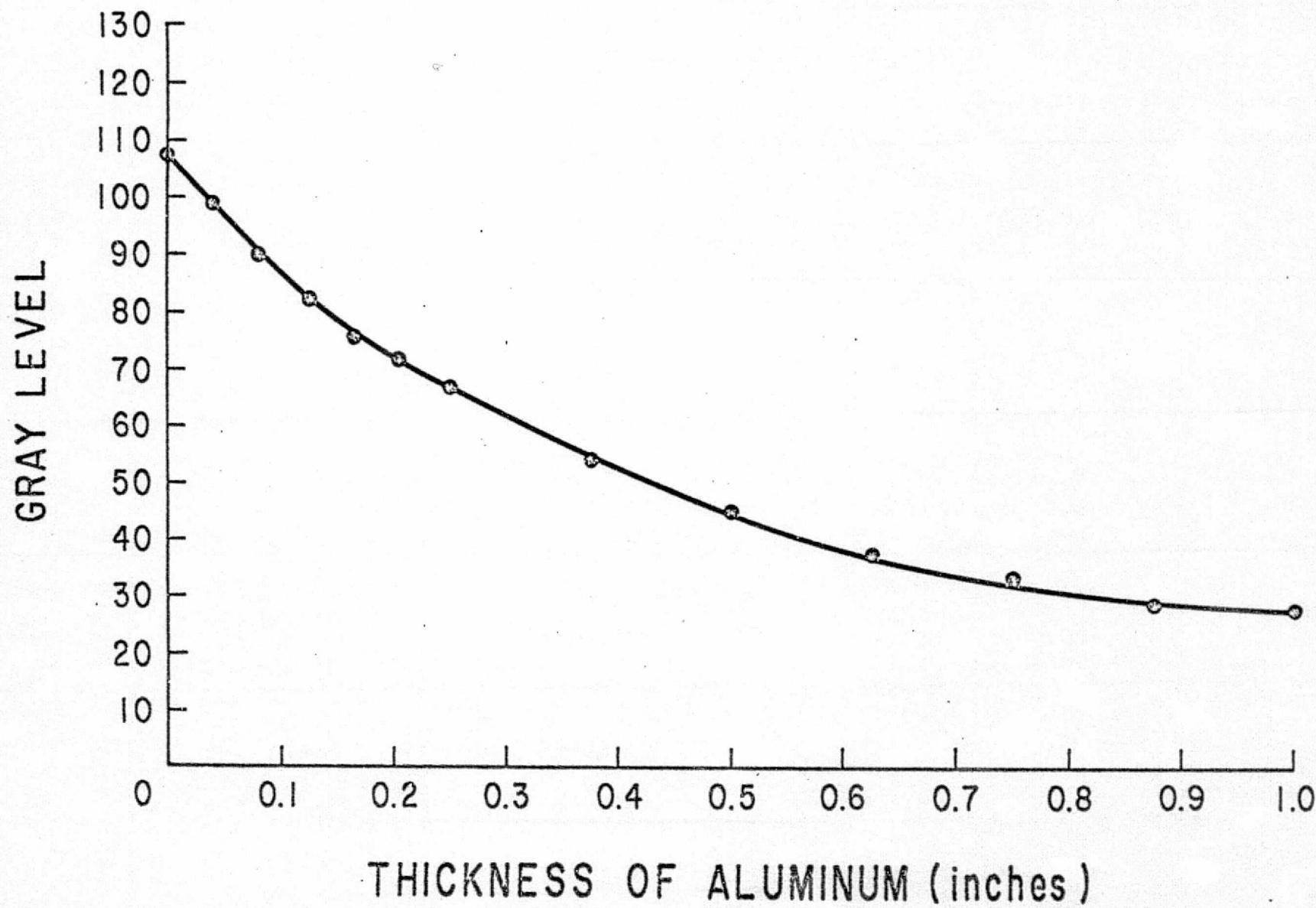
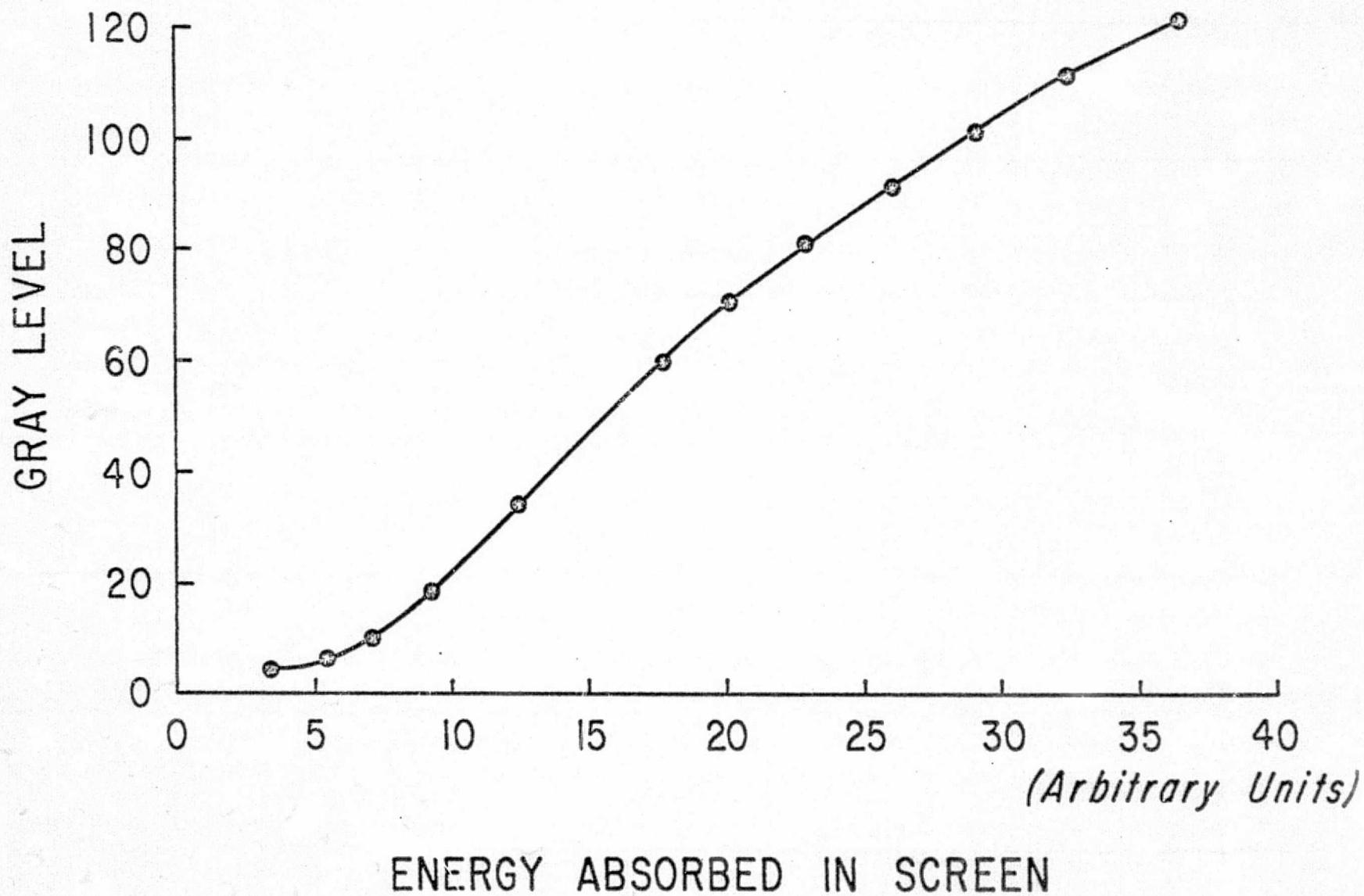


FIG. 4



APPENDIX B

THE CAPABILITY OF FLUOROSCOPIC SYSTEMS FOR THE PRODUCTION OF COMPUTERIZED AXIAL TOMOGRAMS*

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Abstract

Previous work in our laboratories and at other institutions has shown that fluoroscopic images recorded on a video disc can be used successfully for producing computerized-axial-tomograms. The work described in this paper gives a quantitative analysis of the capabilities of such imaging systems, in conjunction with a particular method of data processing, for detecting and imaging changes in object absorptivity. Relations between the degree of contrast or absorptivity and object size required by this type of system can be inferred from the data.

Key Words

Tomography, fluoroscopy, computerized, transaxial, electronic radiography.

Introduction

Work previously done in our laboratories indicated that data derived from fluoroscopic images might be capable of producing computerized axial tomograms.^{1,3,6} In our earliest trials, a phantom made of lucite which contained two 1/4" holes was constructed. The holes were filled with a 1.5% solution of Hypaque. This lucite block was then mounted on a 1 mm thick aluminum sheet which was attached to a protractor capable of being rotated about its center. This assembly was then placed midway between an x-ray tube and the surface of the fluoroscopic input screen of an image intensifier with care to keep its center along the central ray of the x-ray beam as close as could be determined by mechanical means.

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Sixty images spaced 3° apart were recorded, digitized, and processed to reduce the statistically independent noise. The results obtained showed that for an attenuation contrast $(\frac{\Delta I}{I} \times 100)$ of 23%, objects as small as 1 mm could be satisfactorily imaged and also that the approximately 6 mm columns of Hypaque having an attenuation contrast of approximately 12% were also easily imaged.⁸

These preliminary results and the results of Robb, et al.,¹¹ were so promising that we proceeded with our investigations. As a next step we placed a freshly sacrificed medium size rat into a lucite tube having a 2" outside diameter and a 1/8" wall thickness. Video images (180, 1° apart) were obtained using a C-arm x-ray unit having a Philips 6" image intensifier for its fluoroscopic unit. The rat's heart was localized under fluoroscopy. The midplane was marked by a lead arrow. This level was later used as the center of a vertical 8-TV line band for the digitized density matrix. The reconstructed section therefore represented a thickness of 2 mm (in the rat). These results were recently presented at a conference on cardiovascular imaging and will be published in the proceedings.⁴ Ribs, sternum, and spine were clearly demonstrated. The heart chambers were visualized even though no contrast agent whatsoever was used. Most remarkable was the clear presentation of the spinal canal.

In view of these facts we felt a more comprehensive and quantitative evaluation of this type of system for producing computerized-transaxial-tomograms was warranted.

Method

A phantom similar in construction, but containing greater detail, to that used by McCullough, et al.,⁹ was constructed. This phantom contained a cylindrical lucite block placed so that it could be rotated within a rectangular block of lucite. The air gap surrounding the phantom was approximately 1 mm. Rods of various materials and diameters were embedded within the disk. The inclusion of the rectangular block allowed us to use a greater portion of the video signal for the desired gray scale range since this resulted in a more uniform x-ray intensity incident across the entire portion of the fluoroscopic screen. Since we do not have our system assembled into a scanner, and it is more convenient, although more demanding on geometric alignment, to rotate the object rather

than the source and imaging device, we attached a protractor to the test cylinder for this purpose. This assembly is shown in Fig. 1. The details of its construction are shown in Fig. 2. The diameter of the large lucite cylinder was 3.4" and its thickness was 0.75". The x-ray beam was restricted by collimation to an incident field size slightly less than the vertical dimension (0.75").

The target-to-screen distance was 50" and the center of the phantom-to-screen distance, 23.5". All images were recorded at 100 kVp. The effective potential was determined to be 42 keV from the slope of an absorption curve using lucite absorbers. 180 images spaced 1° apart were recorded on a video disc. The preliminary processing performed on the digitized images was as described in references 3 and 7. This resulted in one hundred and eighty 1×80 matrices which were then used in the reconstruction algorithm.⁸ Normalization to account for changes in x-ray flux was accomplished by using an average value of the gray scale derived from an area outside the cylinder where the lucite pathlength remained constant (without perturbations due to test rods).

In affecting the reconstruction, use was made of the fact that the video signal level is proportional to the product of the absorption coefficient and the pathlength of the absorber placed in the beam,⁵ over much of the operating range of these systems. All recordings were made using beam intensities which placed the video signal levels in a range where this had been verified experimentally. In this investigation the recordings were made from images formed on our large screen, low light level TV system.²

The reconstruction was accomplished using a convolutional technique. Kak,⁸ has shown that for beams with divergences up to approximately 20° one can use the convolution algorithm designed for the parallel radiation beam with negligible degradation, both quantitatively and from the point of view of visual quality. An exact convolutional algorithm for such a fan beam configuration is now available.¹²

Results

A reconstructed image is shown in Fig. 3. The light areas represent objects having lower absorptivity than the surrounding lucite, while the dark areas depict the objects with higher absorptivity than the surrounding lucite. In addition, the bright ring represents the

air gap introduced by the spacing between the disk and block. The pertinent data for each rod embedded in the lucite cylinder is given in Table I.

In calculating the values given in this table, $\Delta\mu$ was taken as the difference between the total linear absorption coefficients for lucite and that of the particular absorber. That is,

$$\Delta\mu = \mu_{\text{lucite}} - \mu_{\text{test material}} \quad (1)$$

The last column was calculated using,

$$\frac{I_{\text{lucite}} - I_{\text{test material}}}{I_{\text{lucite}}} \times 100 = (1 - e^{-\Delta\mu \times d}) \times 100 \quad (2)$$

where the I's represent the values of the attenuated beam for a given thickness of the designated materials.

The computed values have a number of uncertainties associated with them. First, there is the determination of the effective energy. This error will tend to introduce only a minimal uncertainty in the value of $\Delta\mu$ since the energy dependence of the attenuation coefficients of the materials used is quite similar. Second, the values of the attenuation coefficients used were taken from NBS-Circular 583,¹⁰ which were derived from theoretical calculations. The author of this circular gives an overall estimate of the error as approaching 10 percent for energies below 50 keV. Third, additional uncertainties were introduced since values for the effective energy (42 keV) were arrived at by a graphical interpolation in energy. Fourth, the values for fluorine required an additional graphical interpolation in atomic number. In view of these, it is not unreasonable to assign an error to the differential absorptions of ± 0.6 for the largest diameter, with values as low as ± 0.04 for the smallest diameter. These are based on an uncertainty of ± 0.005 for all values of $\Delta\mu$.

Of the fifteen rods only three failed to image. These were the 1.6 mm polystyrene, the 5.6 mm and 9.5 mm lexan rods. These had differential absorptivities of less than 1% with respect to the lucite which they replaced. In view of the uncertainties in the values of the absorption coefficients, it is difficult to put an exact value on the sensitivity which one can attain with this method. However, it is probably safe to say that this methodology has the capability of imaging objects having differential attenuations of 1% or greater. Also, it is quite evident, that at least for contrasty objects, the resolution approaches 1 mm.

We should also like to call to the reader's attention three artifacts which appear in the reconstructed image: First, a bright spot, adjacent to the image of the aluminum rod, which appears on the outer edge of the picture (11 o'clock); second, a dark area which appears at the very center of the picture; third, streaks which appear outside of the area of the reconstruction. At the present time we do not have a satisfactory explanation for these.

Discussion

The results given above were attained using a primary imaging system having noise and resolution characteristics distinctively inferior to most systems presently in use clinically. It is entirely possible that improvements in these characteristics would produce reconstructions demonstrating even greater capabilities.

While the use of fluoroscopic images does not seem to produce better reconstructions than the EMI type scanner, a number of advantages manifest themselves. First, the patient dose per image would be between 0.25 and 1.5 mR/exposure. The exposure at the surface of the phantom used in this investigation was approximately 1 mR/exposure and at the input screen approximately 0.6 mR. Therefore, since the entire fan beam geometry is recorded at the same time, the total patient exposure would be at the highest 0.27 R per scan or as low as 45 mR for 180 recordings. Other results^{8,11} indicate that reasonably accurate reconstructions can be obtained using many less images. This would, of course, result in a further reduction of patient dose.

A scanner constructed on the basis of using fluoroscopic images can be very much mechanically simplified since both source and detector are large and stationary with respect to each other since a fan beam geometry is used and, therefore, all portions of the images are recorded simultaneously. This would also allow one to use much shorter recording times. Recordings can be made in the millisecond range only being limited by the x-ray generator's ability to deliver approximately 5×10^{-2} MAS in the shortest possible time desired. It is also possible, using short exposures, to pulse the x-ray source with a physiological signal so as to record and reconstruct at a particular phase of some cyclical phenomena.

Another aspect which is under investigation is the ability to reconstruct multiple tomographic sections from one set of recorded images. This has apparently been achieved by Robb, et al,¹¹ although it is not apparent from theoretical considerations that loss of resolution will not be objectionable. However, should this be possible, a significant reduction in patient dose would result and, of course, the total time per examination would be greatly reduced.

Another aspect which requires further investigation is the use of a grid. The use of a grid should improve image contrast through the removal of scatter. The resulting increase in patient dose would have to be weighed against the resulting improvement in reconstruction sensitivity.

Conclusion

1. The use of video images recorded from fluoroscopic x-ray systems of the type normally used in clinical radiology or those derived from large screen low-light level TV systems can be used to generate satisfactory computerized transaxial tomograms.
2. The capability of such systems to image low contrast objects is of the order of 1% in differential attenuation.
3. The limiting resolution of this type of system is between 1 and 2 mm.
4. Using single scan TV recording and a pulsed x-ray source, reconstructions gated by physiological signals can be achieved.

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TABLE I

Relative absorption of rods embedded in lucite test disc, at 42 keV.

Material	$\Delta \mu$ (cm ⁻¹)	diameter (cm)	$(1 - e^{-\Delta \mu \times d}) \times 100$
Al	- 1.12	0.08	9 ± 0.04
		0.16	20 ± 0.08
Polystyrene CH	0.041	0.16	0.7 ± 0.08
		0.56	2.3 ± 0.3
		0.95	3.8 ± 0.5
		1.27	5.1 ± 0.6
Paraffin CH ₂	0.062	0.56	3.4 ± 0.3
		0.95	5.7 ± 0.5
		1.27	7.6 ± 0.6
Teflon C ₂ F ₄	- 0.297	0.16	5.0 ± 0.08
		0.32	10 ± 0.2
		0.56	18 ± 0.3
Lexan C ₁₆ H ₁₄ O ₃	5 × 10 ⁻³	0.56	0.3 ± 0.3
		0.95	0.5 ± 0.5
		1.27	0.6 ± 0.6

Figure Captions

Fig. 1: Test phantom assembly.

Fig. 2: Scale drawing of test phantom details.

Material	Diameter (mm)
A. Polystyrene	12.7
B. Paraffin	12.7
C. Teflon	1.6
D. Polystyrene	9.5
E. Polystyrene	1.6
F. Lexan	9.5
G. Aluminum	1.6
H. Paraffin	9.5
I. Teflon	3.2
J. Polystyrene	5.6
K. Paraffin	5.6
L. Lexan	5.6
M. Aluminum	0.8
N. Teflon	5.6
O. Lexan	12.7

Fig. 3: Computer reconstruction (transverse-axial) of the phantom shown in Fig. 2. Light areas are those having a lesser absorption than that of lucite, while the dark areas indicate objects having greater absorption than the lucite.

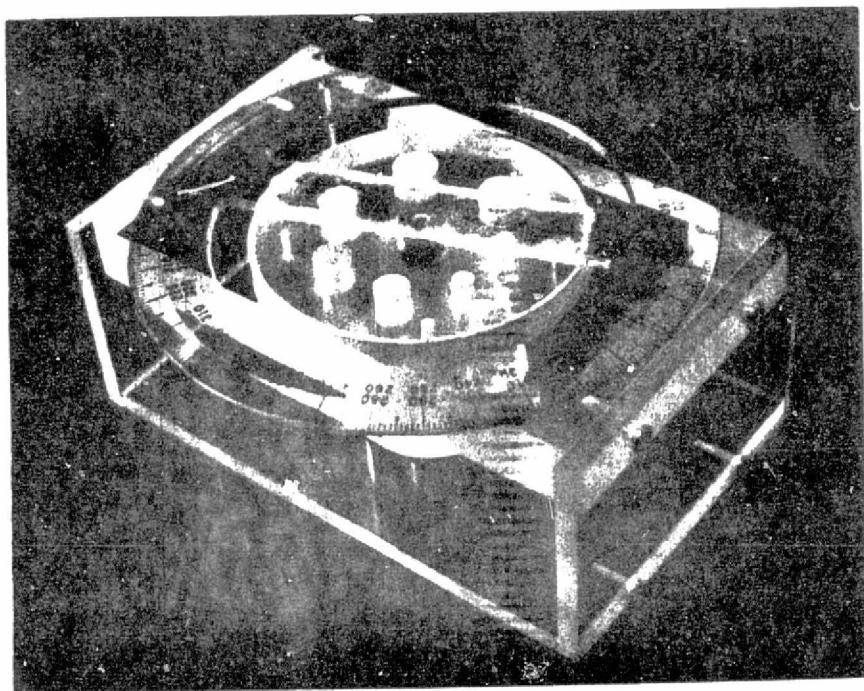


FIG. 1

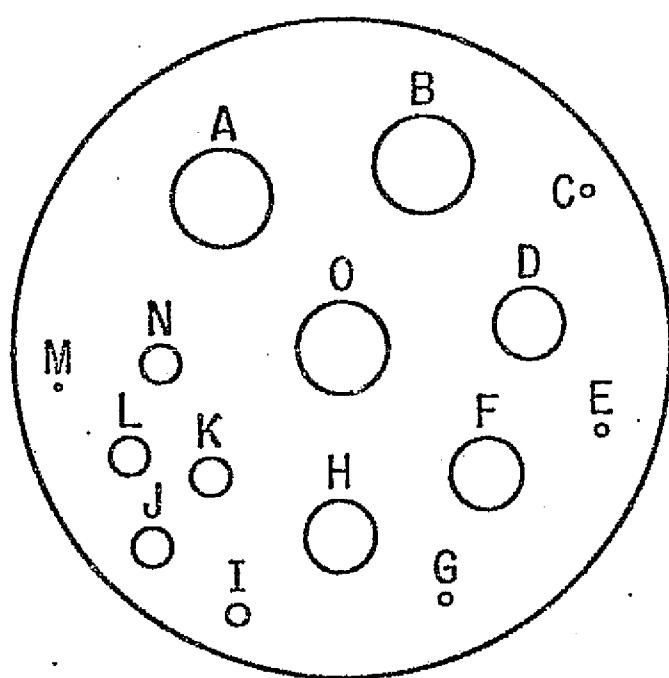


Fig. 2
Baily et al

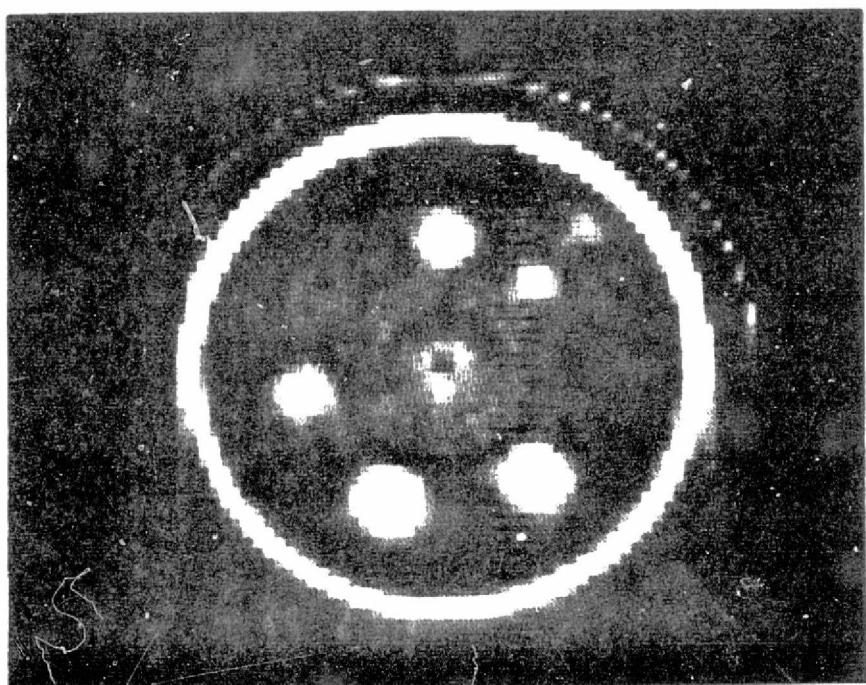


FIG. 3